The Biomechanics of Knees at High Flexion Angles Before and After Total Knee Arthroplasty

by

Ephrat Most

SM Mechanical Engineering
Massachusetts Institute of Technology, 2000

Submitted to the Department of Mechanical Engineering
in Partial Fulfillment of the Requirements for the Degree of
Doctor of Science in Mechanical Engineering
at the

MASSACHUSETTS INSTITUTE OF TECHNOLOGY

~~February~~ JUNE 2004

Author...
Department of Mechanical Engineering
February 29, 2004

Certified by...
Guoan Li
Assistant Professor of Orthopedic Surgery
Harvard Medical School
Research Supervisor

Certified by...
Derek Rowell
Professor of Mechanical Engineering
Thesis Supervisor

Accepted by........... ....
Ain A. Sonin
Chairman, Department Committee on Graduate Students

The Biomechanics of Knees at High Flexion Angles Before and After Total Knee Arthroplasty

by

Ephrat Most

Submitted to the Department of Mechanical Engineering
in partial fulfillment of the requirements for the degree of
Doctor of Science in Mechanical Engineering

Abstract

Total Knee Arthroplasty (TKA) was initially developed to alleviate pain in the case of severe arthritis of the knee. Restoration of knee motion has been an on going issue for the last decade. Contemporary TKAs appear to provide good knee function in the range of zero to 120° of flexion for most patients. However, many patients rarely can flex tier knees beyond 120° after TKA.

Limited information is available regarding the biomechanics of the knee beyond 120° of flexion. Little is known about the biomechanical function of the posterior cruciate ligament in cruciate retaining TKA designs and the interaction of the cam-spine mechanism in posterior-stabilized TKA designs at flexion angles greater than 120°. The role of soft tissue constraint at high flexion angles has not yet been explored.

The objective of this work was to investigate the biomechanics of the knee at high flexion angles before and after TKA. An *in vitro* experimental robotic set-up was used to measure six degrees-of-freedom kinematics and soft tissue kinetics of the intact knee. Contemporary TKA designs were then tested on the same specimen using this system to examine the limitations of currently available components to achieve high knee flexion. Both passive and muscle load kinematics were examined. Femoral translation and tibial rotation of the reconstructed knees were compared with that of the intact knees from full extension to 150° of flexion.

The study showed that in the intact knee, the amount of posterior femoral translation increased with increasing flexion angles on the passive path and under simulated muscle loads. Similar trend was noted for all TKAs. Yet, after any TKA, the knee exhibited a reduction in posterior femoral translation relative to the intact knee. The posterior cruciate ligament in all knees carried lower load at high flexion as compared to the peak load it carried at mid knee flexion. The engagement of the femoral cam with the polyethylene spine in a posterior-stabilized TKA was correlated with an increasing posterior femoral translation. The function of the menisci was not simulated by any of the TKAs. In all knees, the compression of the posterior soft tissue at high knee flexion was correlated with an increase of posterior femoral translation. It is proposed that posterior femoral translation and internal tibial rotation ate high knee flexion are

necessary but not sufficient features in achieving high knee flexion. Factors such as posterior soft tissue compression and contact mechanics should be considered.

Thesis Chair: Derek Rowell, PhD
Title: Professor of Mechanical Engineering
Massachusetts Institute of Technology

Thesis Supervisor: Guoan Li, PhD
Title: Assistant Professor of Orthopedic Surgery
Bioengineering Laboratory
Massachusetts General Hospital
Harvard Medical School

Thesis Committee: Harry E. Rubash, MD
Title: Chief, Orthopedic Surgery
Massachusetts General Hospital
Harvard Medical School

Thesis Committee: Peter TC So, PhD
Title: Associate Professor of Mechanical Engineering
Massachusetts Institute of Technology

To my family:

Nathaniel (ז"ל) and Ruth Most

Isaac (ז"ל) and Shulamit (ז"ל) Rubinfeld

Eliahu (ז"ל) and Riva (ז"ל) Most

Liora Avigdori

Ayala Most

Sheldon F. Oppenehim

Asking questions is men's finest quality.

- Soloman ibn Gabirol

Acknowledgments

There are quite a few people who I must thank for assisting, guiding, and supporting me throughout the development of this work and my graduate life at MIT. Without their help I would not have been able to complete this work.

First and foremost, I want to thank my advisors and my committee members (alphabetically): Drs. Guoan Li, Derek Rowell, Harry Rubash, and Peter So. It has been an honor and a privilege to work with such knowledgeable and experienced faculty. They constantly challenged me while at the same time encouraging me. In particular, I thank Dr. Li who spent hours and hours teaching me biomechanics. Your knowledge, motivation, encouragement, and advice will be a role model in my career. I express my gratitude to Drs. Rowell and So for guiding me through my MIT experience, offering insight in system development, and providing me with thesis guidance. Professor Rowell, thank you for simply being there for me whenever I needed. Dr. Rubash, I want to thank you for allowing me to work with you on this exciting research. Dr. and Mrs. Rubash, I cannot thank you enough for always finding time for Sheldon and me in your extremely busy schedule.

I also need to thank a number of orthopedic surgeons who assisted me throughout this research, provided me with the clinical aspect of the work, and thought me the secrets of knee replacements: Drs. Eric Otterberg, Shay Zayontz, Steve Schule, Peter Sultan, Sang-Eun Park, and Reuben Gobeize. I hope our paths will cross again.

I must also thank two special people who I closely worked on these various projects: Todd Johnson and Lingga Tanamal from Zimmer. It was a pleasure working with such knowledgeable engineers.

This work could not be done without the assistance of many other people. Rohan Abeyaratne, Leslie Regan, Joan Kravit, and Lori Humphrey, Paula Cohen, and Druan Jedry; I could not have done it without you! Thank you for all your help, guidance and support throughout these past 5.5 years. You have a special place in my heart.

I thank the members of the Bioengineering Laboratory at Massachusetts general Hospital: Janine (ladies first), Lou, Jeremy, Ramp, Thomas, Jeremie, and George who have been there for me through out many of my experiments and moods and provided me with everlasting friendships. I thank you all for your energetic spirit and keen insights.

I would like to thank Maria, Sven and Katherina (Ticki) Scheffler who have been my second family. Thank you for all the advice, support, guidance, and just for being there when I needed.

As part of my research involved testing of human cadavers, I want to thank the donors and their families for believing in improving the human conditions. Your generosity has been a tremendous contribution to knee biomechanics research.

Last but not least, I want to thank my family for all their support and sacrifices throughout the years. From the bottom of my heart I thank my mother, Ruth Most, for giving me the best I can get, often, more than I deserved. To my sisters, Liora (Lulu) and Ayala (Ayal) for helping mom when I was away and keeping me up-to-date with all the gossip over in Israel. To my husband, Sheldon: there are not enough words and space to express my thanks to you. You have been the light of my life. I could not have done it without you!

To all of these people, and more, I cannot express the thanks and gratitude that is in my heart.

I remember with love, the first engineer in my life, my father Nathaniel Most (ז"ל). I walk in your shoes.

This work was funded by the National Science Foundation (NSF), Zimmer Inc., and Hugh Hampton Young Memorial Fund Fellowship (MIT). TKA components were generously provided by Zimmer.

Ephrat (Ephi) Most.

Contents

List of Figures

List of Tables

Chapter 1

INTRODUCTION

1.1 Motivations and Objectives

Approximately 16 million Americans suffer from osteoarthritis (OA) and more than 5 million American adults suffer from knee OA (American Academy of Orthopaedic Surgeons). OA is characterized by slowly progressive cartilage degeneration, a thickening of the subchondral bone, the formation of osteophytes, and bone changes [1]. Patients with OA often suffer from pain and joint stiffness. In the case of severe OA, the disease can be debilitating by limiting patients from performing daily living activities as a result of severe reduction in joint range of motion. Total knee arthroplasty (TKA) was initially developed to alleviate pain in cases of severe OA of the knee and allowed patients to return to their daily activities. As the procedure evolved, the long-term success rate in terms of patient satisfaction has improved to in excess of 85% at 10-15 years follow-up [2-6]. Knee range of motion after TKA is considered an important variable in determining clinical outcome [7, 8]. Yet, the accomplishments achieved in restoration of knee motion and kinematics have not paralleled those of pain relief [9].

Attempts at improving knee flexion after TKA have remained of great interest among researchers.

The amount of knee flexion has been linked to functional outcome and activities of daily living [10]. For activities such as sitting on a chair, walking, and stair climbing, the required knee range of motion is limited to approximately -10° to 100°. However, knee flexion beyond 90° is essential and desired in many other day to day circumstances [8]. Rowe et al [11, 12] reported that getting into a bath requires an average knee flexion of 123.3° ± 14.1° and getting out of a tub requires an average knee flexion of 131.3° ± 14.3°. An individual typically needs between 111° and 165° of knee flexion in order to squat, kneel, and sit cross-legged [10, 13, 14]. Weiss et al [15] surveyed 367 post TKA patients as to their interest in participating in sport activities after the surgery. The survey reveled that patients are interested in stretching exercise (56%), kneeling (52%) and gardening (50%) activities; yet patients find it very difficult to perform these activities after TKA. Low impact activities such as gardening, meditation, yoga, golf or a catcher in a softball game, often require knee flexion beyond 150° (Figure 1) [9, 10]. In fact, "golf is a frequent form of exercise for the older population in whom TKAs are usually performed" [16] and to check the line for a putt on the green in golf requires that the individual be able to squat, which requires knee flexion of greater than 120° [9].

Figure 1. Selected examples of low impact activities that require high knee flexion: A. yoga and meditation, B. gardening, and C. catcher during a softball game.

The need of patients to return to their daily living activities after TKA may vary from person to person. In many Western countries, patients' choice in engaging in activities that require high knee flexion (>120°) is considered an additional benefit as these activities are not crucial to the patients' life. However, professions such as construction workers, farmers [17], and plumbers, highly depend on their ability to continue to work after TKA as a source of income. Patients whose movement is restricted after TKA may experience disability that would have both personal and economic consequences. The dilemma of limited knee flexion after TKA may be more apparent in the Far and Middle East countries as many activities that require knee flexion beyond 120° are vital for the patients' daily function in these countries. For example, Japanese and Indian people squat in order to use "Eastern style toilets" [10]. Kneeling is a commonly used position for sitting, praying, or dining [10, 18]. Without the ability to flex beyond 120°, these activities will not be achievable and these patients' life will deviate from their cultural norm.

Although the human knee is capable of flexing more than 150°, contemporary TKAs rarely result in knee flexion greater than 120° (Table 1) [8, 19-28]. The success of any total knee replacement system may be, in part, linked to its ability to optimally restore intact knee function. Current prosthetic designs and surgical techniques may not meet the needs of patients who require high knee flexion for their daily activities. Several arthroplasty designs are available that incorporate modifications aimed at improving the post-operative knee range of flexion [29-31]. However, limited data are available on their function and potential advantages.

Table 1. Reported range of flexion for various TKA designs [8] (Table adopted with permission from Lippincott Williams & Wilkins).

Study	Follow-up (yr)	Design	No. of knees	Mean flexion (degrees) (range)
Cruciate sacrificing				
Goldberg et al.	9	Total condylar	109	95 (15–115)
Insall et al.	6.5	Total condylar	100	89 (no range reported)
Ranawat et al.	13.2	Total condylar	62	99 (65–120)
Cruciate substituting				
Aglietti et al.	5.5	Insall-Burstein	73	96 (70–120)
Emmerson et al.	12.7	Kinematic stabilizer	109	98 (25–130)
Ranawat et al.	4.8	Press-Fit condylar	125	111 (75–135)
Cruciate retaining				
Dennis et al.	11	Cruciate condylar	42	104 (76–120)
Lee et al.	9	Cruciate condylar	144	106 (no range reported)
Malkani et al.	10	Kinematic condylar	119	105 (±11)
Rosenberg et al.	3.5	Miller-Galante	116	105 (45–140)

As part of my Master's thesis, a robotic testing system was developed to investigate joint kinematics [32]. Since then, *in vitro* experimental models incorporating the robotic system have been used to investigate the biomechanics of the intact knee and various TKA designs throughout the entire range of flexion (full extension to 150° of flexion) [33-38]. Many of these experiments are described in detailed in this dissertation.

The overall goal of this work is to better understand the factors that limit the human knee joint from achieving higher degrees of flexion after TKA. In particular, the past several years have been utilized to answer the following questions:

1. How does the intact knee behave at high flexion angels (>120°)? What makes the knee so stable at extreme flexion angles? Which structure(s) guide(s) intact knee motion beyond 120°?

2. What is the role of the soft tissue structure (including posterior cruciate ligament (PCL), menisci, muscles, posterior soft tissue compression) on intact knee motion at high flexion angles?

3. Do contemporary TKA designs restore intact knee kinematics (femoral translation and tibial rotation) from full extension (0°) to full flexion (150°)? If not, to what extend do they restore the kinematics?

4. What is the function of the PCL in a cruciate retaining TKA, particularly at high knee flexion?

5. What is the function of the cam-spine in a posterior-stabilized TKA, particularly at high knee flexion?

6. What is the effect of soft tissue compression on the motion of the reconstructed knee(s) at flexion angles beyond 120°?

7. Is kinematics data sufficient to describe the performance of a given TKA design? How does contact mechanics, predominantly at high knee flexion, comes into play?

8. What happens to the flexion gap with increasing flexion angles to both the intact and the reconstructed knees?

In this work, flexion angles beyond 120° are considered to be high flexion angles. The robotic model in conjunction with clinical studies provides an understanding of the limitations of contemporary knee designs in achieving higher degrees of knee flexion.

This may lead to the refinement of existing designs and development of newer prostheses that will enhance the range of flexion that is achievable following TKA.

1.2 Organization

The work in this dissertation is divided into two main sections. The first deals with the understanding of intact knee kinematics and kinetics throughout the full range of motion (0° - 150°). The second investigates various conventional TKA designs and compares their performance to that of the intact knee.

The text is organized sequentially. Chapter 2 describes the steps taken to adjust the robotic testing system for testing intact and reconstructed knees on the same specimen. Chapter 3 reports a description of different coordinate systems and their effect on the kinematics. This is a critical chapter as it provides the reader with a comprehensive background to better understand the differences that exist in the literature regarding the different studies. Chapter 4 includes a detailed analysis of the intact knee biomechanics from full extension to 150° of flexion. In addition, a literature review of relevant work both on intact knee kinematics and ligament force is presented. In Chapters 5 and 6, the reader can find a description of the biomechanics of two contemporary TKA designs and their compression in relation to the intact knee. Chapter 7 includes a preliminary analysis of the flexion gap in high knee flexion. The flexion gap is thought to be a limiting factor in allowing high knee flexion. The final chapter presents the overall results of this project as well as suggestions for future directions in TKA design and performance.

The dissertation is based on the following papers:

1. "Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads", Li G, Zayontz S, Most E, Otterberg E, Sabbag K, Rubash HE., J Arthroplasty. 2001 Dec;16(8 Suppl 1):150-156.
2. "Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study", Li G, Most E, Otterberg E, Sabbag K, Zayontz S, Johnson T, Rubash H, Clin Orthop. 2002 Nov;(404):214-225.
3. "Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty", Most E, Zayontz S, Li G, Otterberg E, Sabbag K, Rubash HE, Clin Orthop. 2003 May;(410):101-113.
4. "The Kinematics of Fixed- and Mobile-Bearing Total Knee Arthroplasty", Most E, Li G, Schule S, Sultan P, Park SE, Zayontz S, Rubash H, Clin Orthop. 2003 Nov; (416):197-207.
5. "Optimizing Flexion after Total Knee Arthropaslty: Advances in Prosthetic Design", Sultan P, Most E, Li G, Rubash E, Clin Orthop. 2003 Nov; (416):167-173.
6. "Effect of the Posterior Cruciate Ligament on Posterior Stability of the Knee in High Flexion" G. Li, E. Most, L. E. DeFrate, J. F. Suggs, T. J. Gill, H. E. Rubash, Submitted to JBJS May 2003.
7. "Kinematics of the knee at high flexion angles: an in vitro investigation", G. Li, S. Zayontz , L. E. DeFrate, E. Most, J. F. Suggs, H. E. Rubash, Journal of Orthopaedic Research, 2004, Jan 22(1): 90-95.

8. "An Evaluation of Knee Kinematics with a High Flexion Posterior-Stabilized Total Knee Arthroplasty Using an *in vitro* Robotic Experimental System" G. Li; E. Most; P. Sultan; S. Schule; S. Zayontz.; H. Rubash, Submitted to JBJS July 2003.

9. "Sensitivity of the Knee Joint Kinematics Calculation to Selection of Flexion Axes", E Most; J Axe; H Rubash; G Li, Submitted to J biomech August 2003.

10. "Kinematics Analysis of a Conventional and a High-Flexion Cruciate Retaining Total Knee Arthroplasties: An In-vitro Investigation", Most E, Li G, Sultan P, Park SE, Rubash HE, Submitted to J Arthroplasty.

11. "A Comparison of the Contact Behavior between a Conventional and a High-Flexion Cruciate Retaining Total Knee Arthroplasty", Most E, Li G, Papannagari, R, Sultan P, Park SE, Rubash HE, to be submitted.

1.3 References

1. Silver, F., G. Bradica, and A. Tria, *Structure and Biomechanics of Articular Cartilage*, in *The Adult Knee*, J. Callaghan, et al., Editors. 2003, Lippincott Williams & Wilkins: Philadelphia. p. 105-122.
2. Duffy, G.P., R.T. Trousdale, and M.J. Stuart, *Total knee arthroplasty in patients 55 years old or younger. 10- to 17-year results.* Clin Orthop, 1998(356): p. 22-7.
3. Gill, G.S. and A.B. Joshi, *Long-term results of retention of the posterior cruciate ligament in total knee replacement in rheumatoid arthritis.* J Bone Joint Surg Br, 2001. 83(4): p. 510-2.
4. Meding, J.B., M.A. Ritter, and P.M. Faris, *Total knee arthroplasty with 4.4 mm of tibial polyethylene: 10-year followup.* Clin Orthop, 2001(388): p. 112-7.
5. Pavone, V., et al., *Total condylar knee arthroplasty: a long-term followup.* Clin Orthop, 2001(388): p. 18-25.
6. Schai, P.A., T.S. Thornhill, and R.D. Scott, *Total knee arthroplasty with the PFC system. Results at a minimum of ten years and survivorship analysis.* J Bone Joint Surg Br, 1998. 80(5): p. 850-8.
7. Maloney, W.J. and D.J. Schurman, *The effects of implant design on range of motion after total knee arthroplasty. Total condylar versus posterior stabilized total condylar designs.* Clin Orthop, 1992(278): p. 147-52.
8. Li, G., et al., *Improving flexion in total knee arthroplasty*, in *The Adult Knee*, J. Callaghan, et al., Editors. 2002, Lippincott Williams & Wilkins: Philadelphia. p. 1233-1244.

9. Sultan, P., et al., *Optimizing Flexion After Total Knee Arthroplasty: Advances in Prosthetic Design.* Clin Orthop, 2003.

10. Mulholland, S.J. and U.P. Wyss, *Activities of daily living in non-Western cultures: range of motion requirements for hip and knee joint implants.* Int J Rehabil Res, 2001. 24(3): p. 191-8.

11. Rowe, P.J., et al., *Knee joint kinematics in gait and other functional activities measured using flexible electrogoniometry: how much knee motion is sufficient for normal daily life?* Gait Posture, 2000. 12(2): p. 143-55.

12. Myles, C.M., et al., *Knee joint functional range of movement prior to and following total knee arthroplasty measured using flexible electrogoniometry.* Gait Posture, 2002. 16(1): p. 46-54.

13. Hefzy, M.S., B.P. Kelly, and T.D. Cooke, *Kinematics of the knee joint in deep flexion: a radiographic assessment.* Med Eng Phys, 1998. 20(4): p. 302-7.

14. Nagura, T., et al., *Mechanical loads at the knee joint during deep flexion.* J Orthop Res, 2002. 20(4): p. 881-6.

15. Weiss, J.M., et al., *What functional activities are important to patients with knee replacements?* Clin Orthop, 2002(404): p. 172-88.

16. Mallon, W.J. and J.J. Callaghan, *Total knee arthroplasty in active golfers.* J Arthroplasty, 1993. 8(3): p. 299-306.

17. Trousdale, R., et al., *Case Challenges in Hip & Knee Surgery: Knee Challenges: What Would You Do?* Orthopedics, 2002. 25(9).

18. Koshino, T., et al., *Increase in range of knee motion to obtain floor sitting after high tibial osteotomy for osteoarthritis.* Knee, 2002. 9(3): p. 189-96.

19. Aglietti, P., R. Buzzi, and A. Gaudenzi, *Patellofemoral functional results and complications with the posterior stabilized total condylar knee prosthesis.* J Arthroplasty, 1988. 3(1): p. 17-25.

20. Dennis, D., et al., *Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation.* Clin Orthop, 1992. 281: p. 168-176.

21. Emmerson, K.P., C.G. Moran, and I.M. Pinder, *Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up.* J Bone Joint Surg Br, 1996. 78(3): p. 441-5.

22. Goldberg, V.M., et al., *Use of a total condylar knee prosthesis for treatment of osteoarthritis and rheumatoid arthritis. Long-term results.* J Bone Joint Surg Am, 1988. 70(6): p. 802-11.

23. Insall, J.N., et al., *The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements.* J Bone Joint Surg Am, 1983. 65(5): p. 619-28.

24. Insall, J.N., et al., *Total knee arthroplasty.* Clin Orthop, 1985(192): p. 13-22.

25. Lee, J.G., et al., *Review of the all-polyethylene tibial component in total knee arthroplasty. A minimum seven-year follow-up period.* Clin Orthop, 1990(260): p. 87-92.

26. Malkani, A.L., et al., *Total knee arthroplasty with the kinematic condylar prosthesis. A ten- year follow-up study.* J Bone Joint Surg Am, 1995. 77(3): p. 423-31.

27. Rand, J.A., *Comparison of metal-backed and all-polyethylene tibial components in cruciate condylar total knee arthroplasty.* J Arthroplasty, 1993. 8(3): p. 307-13.

28. Ranawat, C.S., C.P. Luessenhop, and J.A. Rodriguez, *The press-fit condylar modular total knee system. Four-to-six-year results with a posterior-cruciate-substituting design.* J Bone Joint Surg Am, 1997. 79(3): p. 342-8.

29. Yamazaki, J., et al., *Hy-Flex II total knee system and range of motion.* Arch Orthop Trauma Surg, 2002. 122(3): p. 156-60.

30. Akagi, M., et al., *The Bisurface total knee replacement: a unique design for flexion. Four- to-nine-year follow-up study.* J Bone Joint Surg Am, 2000. 82-A(11): p. 1626-33.

31. Zimmer, I., *NexGen LPS-Flex Design Rationale*: Warsaw, IN.

32. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

33. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.

34. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

35. Most, E., et al., *Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty.* Clin Orthop, 2003(410): p. 101-13.

36. Gill, T.J., et al., *The biomechanical effect of posterior cruciate ligament reconstruction on knee joint function. Kinematic response to simulated muscle loads.* Am J Sports Med, 2003. 31(4): p. 530-6.

37. Li, G., et al., *Determination of optimal graft lengths for posterior cruciate ligament reconstruction--a theoretical analysis.* J Biomech Eng, 2003. 125(2): p. 295-9.

38. Li, G., et al., *Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads--an in vitro experimental study.* J Orthop Res, 2002. 20(4): p. 887-92.

Chapter 2

DEVELOPMENT OF THE ROBOTIC TESTING PROTOCOL FOR STUDYING THE BIOMECHANICS OF INTACT AND RECONSTRUCTED KNEES

2.1 Motivations and Objectives

Total knee arthroplasty (TKA) is performed, in part, to allow patients to return to their pre-operative activities by restoring their normal knee function. Failure to mimic intact knee kinematics is thought to contribute to implant loosening and polyethylene wear, which ultimately lead to revision [1].

Several methods for studying knee kinematics are reported in the literature [2-18]. In vivo kinematics measurements include the use of optical markers, attached to the skin or directly on the bone, to track the relative motion of the tibia with respect to the femur [2-5, 15, 16]. Lafortune [16] described the angular and linear three dimensional knee kinematics using intra-cortical pins fixed directly on the femoral and tibial bones during walking activity. The limiting factors in this study include the limited degree of freedom, no high flexion information, and the inability to directly compare different knee stages on the same patient. Andriacchi et al [2] developed the point cluster technique to measure the six degrees of freedom motion of a knee. The markers are uniformly distributed on the subject's soft tissue limbs (i.e. thigh and shank) and their motion is tracked by an optoelectronic digitizer. Nagura et al [3] used six retro-reflective markers, an opto-electronic system, and a force plate to analyze the mechanical loads in human knee joints exhibited during four different activities, including deep kneeling. They concluded that at high knee flexion, the knee experiences large forces and moments. The limitations of their study include neglecting the contact force between the thigh and the calf, and the inability to directly compare the pre- and post-operative conditions on the same subject.

In vivo investigations provide us with insightful data regarding the motion (kinematics) of the knee either before or after TKA. However, rarely do we know the preoperative kinematics for the same patient undergoing TKA and, therefore, a direct in-vivo comparison between the normal and reconstructed knee cannot be determined. Consequently, it is vital to perform in vitro investigation where by the healthy knee serves as its own control for any TKA performed.

Most in vitro testing systems provide useful information with only one aspect of research such as limited degrees of freedom kinematics, or limited range of flexion.

Often, high knee flexion (>120°) cases are excluded [6-9, 13]. Lewis at al [13] developed a system that measures three dimensional joint motion and ligament forces using a combination of instrumented spatial linkage (ISL) system, buckle transducers, and pneumatic load apparatus. They reported a large anterior cruciate ligament force (90N) at 20 degrees of knee flexion. No information was given in regards to higher flexion angles. Kirstukas, Lewis and Erdman [8, 9] designed a six degrees ISL system to measure joint motion. However, they reported that limited resolution of the devices that monitor the position of the linkage joints exists. The Oxford Knee Rig (OKR) has been used by several groups to investigate the movement of the knee joint and patellofemoral force in six degrees or freedom [10, 17-20]. Wilson et al [17] used fifteen human cadaver knees to study the relative motion of the tibia with respect to the femur for the unloaded knee using the OKR and an electromagnetic tracking system from full extension to 100° of flexion. They reported that with increasing knee flexion, coupled tibial rotation and ab/adduction occurred.

Recently, a robotic testing system was introduced to examine the six degree of freedom knee kinematics [21-26]. Rudy et al [21, 23-25] developed a robotics-based joint testing system that offers the ability to control both the paths of motion as well as the acting forces. The system provides not only the measurement of structural properties, but also the ability to store and repeat the six degrees-of freedom motion under different loading conditions. In response to external loads, the robot can learn the complex motion of the knee specimen and can reproduce these motions in subsequent tests. Their system was used extensively to study intact knee kinematics and ligament forces on the passive path and under muscle load conditions for knee flexion up to 120° [22, 27-32].

Most [26] adopted the robotic testing system and expanded the concept to allow for in-vitro testing of intact and TKA knees on the same specimen. The test system is composed of a six degree-of-freedom robotic manipulator (Kawasaki UZ150®, Kawasaki Heavy Industry, Japan) and a six degree-of-freedom load cell (JR3 DSP-based force sensor receiver, JR3 Inc., Woodland, CA). A control algorithm, written to account for the coupling effects of the different degree-of-freedom of the knee, was developed. Using a personal computer, the robot and the load cell were controlled to allow for both displacement and force modes. Using the system, the kinematics of knees, the tibio-femoral contact, and soft tissue forces can be measured. Detailed description can be found in Most Master's thesis [26]. This chapter provides the principles, specimen installation, and detailed testing protocol used in the different studies through out this dissertation work.

2.2 Coordinate System Establishment

The general term "knee kinematics" usually refers to rotation about and translation along defined coordinate axes [33]. It is therefore critical to define clear and concise coordinate systems to both the tibia and the femur such that the relative motion between the two bones can be quantified. Through the entire development, the femoral and tibial bones are to be considered rigid bodies.

The knee coordinate system is constructed by digitizing (MicroScribe 3DX® Digitizer, Immersion Corporation, San Jose, CA) several anatomic points on the knee. The longitudinal axis of the tibia (x) is built by connecting two points of the tibial shaft parallel to the posterior cortex in the longitudinal direction of the tibia (Figure 2A). The transepicondylar (medial-lateral) axis of the femur (y) is defined as a line connecting the most prominent point within the lateral collateral ligament insertion site and the most prominent anterior to the sulcus point within the insertion site of the medial collateral ligament (Figure 2B) [34, 35]. By taking the cross product of these two axes, the anterior-posterior axis of the knee (z) is created (Figure 2A). The origin of the system (knee center) is chosen as the midpoint of the transepicondylar line (Figure 2B).

The femoral coordinate system and the tibial coordinate system coincide with each other at full extension (initial position) under no load condition. Thus, only one coordinate system is needed to be defined initially.

The knee specimen is aligned so that the load cell can measure three force and three moment components along and about a cartesian coordinate system. As the knee responds to external loads, the tibial coordinate system moves with the tibia. At that point, the coordinate system of the femur no longer coincides with the coordinate system of the tibia. Therefore, the translation vector and the rotation matrix of the tibia with respect to the femur must be evaluated to determine the knee kinematics. An Euler sequence (y-z-x) was adopted to describe the tibial rotation. A derivation of the translation vector and rotation matrix is given in Most [26].

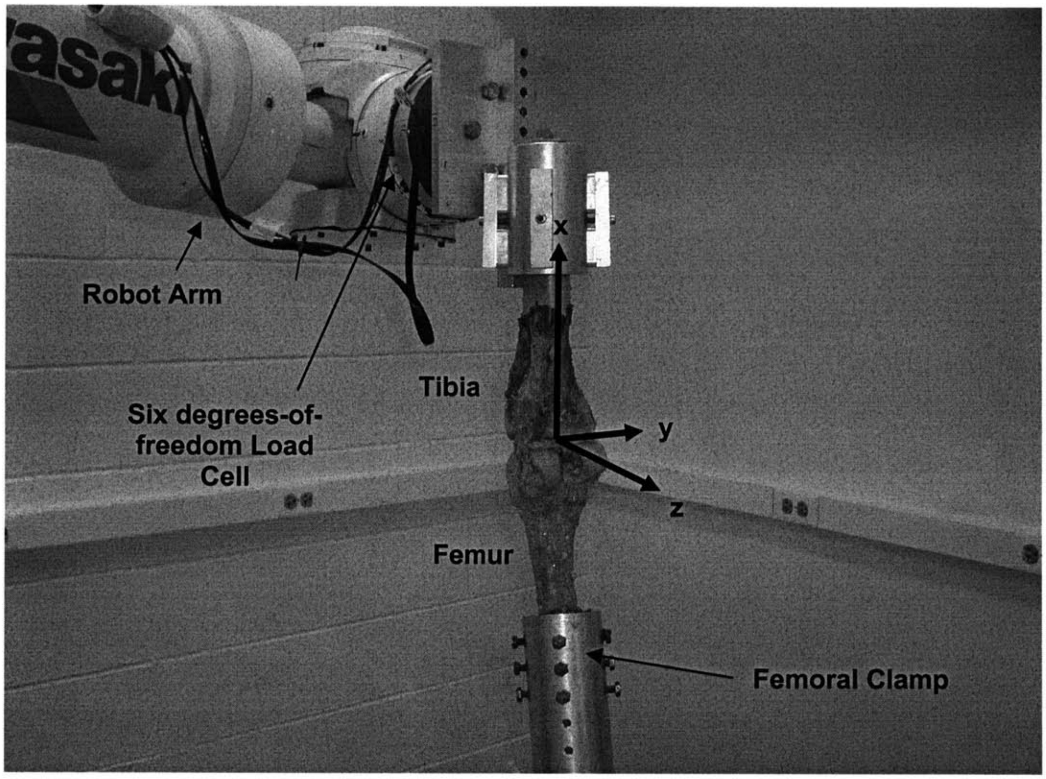


Figure 2A: The figure shows the installation of the specimen and the construction of the coordinate system. For purposes of demonstration only, all soft tissues were removed.

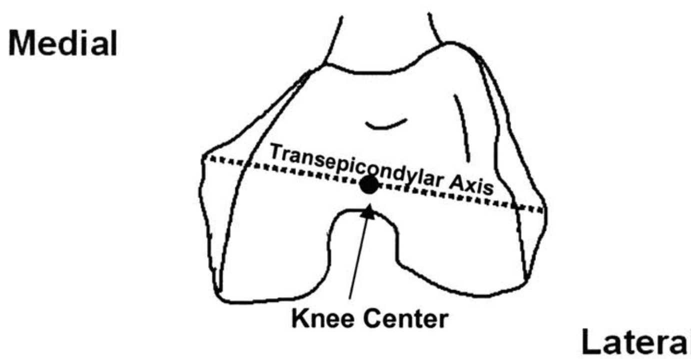


Figure 2B: The figure outlines on an anterior view of a femur and the transepicondylar axis.

2.3 Specimen Preparation and Installation

A minimum of eight post-mortem human knee specimens are used in each study (individual studies are described in upcoming chapters). Prior to testing, each specimen is thawed over night at room temperature. Each knee includes approximately 25cm of bone above and below the joint line, leaving all soft tissues (capsule, ligaments, menisci, posterior capsule, skin, fat, and muscles) around the knee joint intact. The fibula is fixed to the tibia in an anatomical position by a cortical bone screw.

Each knee is manually examined and flexed through its entire range of flexion to assure suitable range of flexion for a given test. Each knee is also x-rayed in both anterior/posterior and medial/lateral views to verify that the knee is healthy with no prior injuries or surgeries. Knees that do not follow these guidelines are excluded from the study.

Both femoral and tibial shaft ends are exposed and potted, using polymethylmethacrylate (PMMA), to enable secured mounting of the specimen on the robotic system. During the experiment, the femur cylinder is rigidly fixed to a specially designed clamp that allows 6 degrees-of-freedom positioning of the femur relative to the robotic system base (Figure 2A). The tibial cylinder is rigidly fixed to the robot arm through the load cell. This set-up allowed the tibia to freely move with the robot arm in 6 degrees-of-freedom about the femur. To avoid dehydration of the specimen, 0.9% saline was regularly sprayed over the specimen.

Each experiment includes the simulation of an unloaded knee state (passive path) as well as loaded knee state (under muscles loads). The later is performed using a pulley and weight system [36]. Prior to each experiment, the tendons of each muscle are manually isolated and a rope is sutured to each one by means of polyester sutures (Ethibond Excel, Ethicon Inc, Johnson & Johnson). During a given experiment, weights are hung from the free end of the rope to simulate muscle forces. Muscular forces of the quadriceps and the medial and lateral hamstrings (semitendinosus/ semimembranosus and biceps femoris) and their co-contraction are simulated in each study.

2.4 Testing Protocol

The robotic testing system allows for multiple tests on the same specimen thereby eliminating inter-specimen variations. A general testing protocol is presented in Figure 3. Prolonged testing of soft tissue may lead to tissue degradation, therefore, only a segment of the protocol can be performed during a given experiment. Throughout this thesis, several experiments were performed to answer all the questions raised in the motivation and objective chapter. In every experiment, the intact knee was tested first and served as the baseline reference for the remainder of the study. A detailed description of each test will be given in the following chapters. Each test includes the determination of the passive path and the kinematics due to the application of various muscle loads.

2.4.1 Passive Path Determination

The passive path is the characteristic behavior of the knee motion determined by the articular geometry and the surrounding soft tissue constraints. In our studies, a passive position at a specific flexion angle is defined as the position of the knee where the knee carries minimal load under no external load. After the knee is mounted onto the joint testing system, a series of passive positions were defined at one-degree increments between 0° and 150° of flexion. At each degree, knee positions for the remaining 5 degrees of freedom were determined such that the residual forces and moments at the knee joint center (midpoint of the transepicondylar line) were minimized (<5 N and 0.5

N-m for normal knees, respectively). These pre-determined passive positions represented the relative position of the tibia with respect to the femur at which the joint carried a minimal load. This series of passive positions formed a *passive path* between full extension to full flexion (150°). This path was then used as the reference position for the subsequent application of simulated physiological loads during testing.

2.4.2 Kinematics under Various Muscle Loads

Simulation of quadriceps and hamstring muscles and their co-contraction was performed using a pulley system, as previously described. Muscular forces of the quadriceps and the medial and lateral hamstrings (semitendinosus/ semimembranosus and biceps femoris) at 2:1 ratio are simulated in each study and at given flexion angles (0°, 30°, 60°, 90°, 120° and 150°). In general, three muscle loads are applied as follows: (1) an isolated quadriceps force of 400 N; (2) a combined quadriceps and hamstrings load (400 N and 200 N, respectively); and (3) an isolated hamstring force of 200 N. When a knee reaches a selected flexion angle along its passive path, muscle loading is then applied. At this point, the tibia is able to move along the remaining 5 degrees of freedom until reaching an equilibrium position where the applied load is balanced by the constraint forces generated inside the knee joint. The new knee position is then recorded by the robotic manipulator. The test is repeated at each selected flexion angle for all the muscle loads.

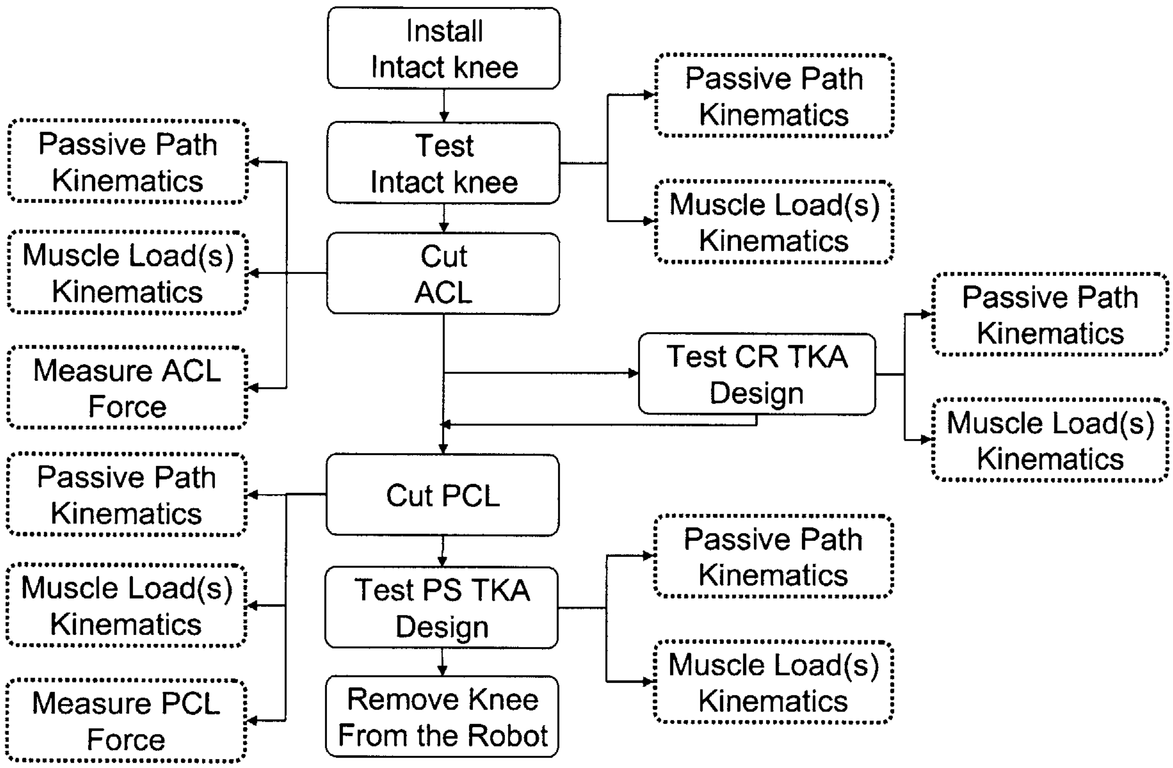


Figure 3: A detailed flow chart describing several scenarios for testing a single healthy specimen using the robotic system. ACL ≡ anterior cruciate ligament; PCL ≡ posterior cruciate ligament; CR ≡ cruciate retaining; PS ≡ posterior stabilized; TKA ≡ total knee arthroplasty.

2.4.3 Ligament and Soft Tissue Force Measurements

Once the passive path (U_{pasive}) and the kinematics under various muscle loads (U_{muscle}) are determined the forces in the different ligaments and soft tissue can be measured by the principle of superposition [21, 22, 26]. The protocol in Figure 3 outlines the procedure. For example, to measure the in situ forces in the anterior cruciate ligament (ACL), the robotic testing system replays the pre-recorded kinematics (U_{pasive} and U_{muscle}) and the load cell system measures the corresponding forces in the ACL ($F1_{ACL\text{-}passive}$ and $F1_{ACL\text{-}muscle}$). The ACL is then resected and the knee kinematics (U_{pasive} and U_{muscle}) are replayed again. The load cell records the forces at the knee center after

ACL resection ($F2_{ACL\text{-}passive}$ and $F2_{ACL\text{-}muscle}$). According to the principle of superposition [21, 22, 26], the difference between the forces measured before and after resecting the ACL represent the ACL forces:

$$\text{ACL forces on the passive path} = F2_{ACL\text{-}passive} - F1_{ACL\text{-}passive}$$

$$\text{ACL forces under combined muscle loads} = F2_{ACL\text{-}muscle} - F1_{ACLmuscle}$$

This procedure can be followed for every ligament and soft tissue as long as only one structure is being measured at a time.

2.5 References

1. Godest, A.C., et al., *A computational model for the prediction of total knee replacement kinematics in the sagittal plane.* J Biomech, 2000. 33(4): p. 435-42.
2. Andriacchi, T.P., et al., *A point cluster method for in vivo motion analysis: applied to a study of knee kinematics.* J Biomech Eng, 1998. 120(6): p. 743-9.
3. Nagura, T., et al., *Mechanical loads at the knee joint during deep flexion.* J Orthop Res, 2002. 20(4): p. 881-6.
4. Alexander, E.J. and T.P. Andriacchi, *Correcting for deformation in skin-based marker systems.* J Biomech, 2001. 34(3): p. 355-61.
5. Lucchetti, L., et al., *Skin movement artefact assessment and compensation in the estimation of knee-joint kinematics.* J Biomech, 1998. 31(11): p. 977-84.
6. Kinzel, G.L., et al., *Measurement of the total motion between two body segments. II. Description of application.* J Biomech, 1972. 5(3): p. 283-93.
7. Kinzel, G.L., A.S. Hall, Jr., and B.M. Hillberry, *Measurement of the total motion between two body segments. I. Analytical development.* J Biomech, 1972. 5(1): p. 93-105.
8. Kirstukas, S.J., J.L. Lewis, and A.G. Erdman, *6R instrumented spatial linkages for anatomical joint motion measurement--Part 1: Design.* J Biomech Eng, 1992. 114(1): p. 92-100.
9. Kirstukas, S.J., J.L. Lewis, and A.G. Erdman, *6R instrumented spatial linkages for anatomical joint motion measurement--Part 2: Calibration.* J Biomech Eng, 1992. 114(1): p. 101-10.

10. D'Lima, D.D., et al., *Comparison between the kinematics of fixed and rotating bearing knee prostheses.* Clin Orthop, 2000(380): p. 151-7.

11. van Dijk, R., R. Huiskes, and G. Selvik, *Roentgen stereophotogrammetric methods for the evaluation of the three dimensional kinematic behaviour and cruciate ligament length patterns of the human knee joint.* J Biomech, 1979. 12(9): p. 727-31.

12. DesJardins, J.D., et al., *The use of a force-controlled dynamic knee simulator to quantify the mechanical performance of total knee replacement designs during functional activity.* J Biomech, 2000. 33(10): p. 1231-42.

13. Lewis, J.L., W.D. Lew, and J. Schmidt, *Description and error evaluation of an in vitro knee joint testing system.* J Biomech Eng, 1988. 110(3): p. 238-48.

14. de Lange, A., R. Huiskes, and J.M. Kauer, *Measurement errors in roentgen-stereophotogrammetric joint-motion analysis.* J Biomech, 1990. 23(3): p. 259-69.

15. Reinschmidt, C., et al., *Effect of skin movement on the analysis of skeletal knee joint motion during running.* J Biomech, 1997. 30(7): p. 729-32.

16. Lafortune, M.A., et al., *Three-dimensional kinematics of the human knee during walking.* J Biomech, 1992. 25(4): p. 347-57.

17. Wilson, D.R., et al., *The components of passive knee movement are coupled to flexion angle.* J Biomech, 2000. 33(4): p. 465-73.

18. Zavatsky, A.B., *A kinematic-freedom analysis of a flexed-knee-stance testing rig.* J Biomech, 1997. 30(3): p. 277-80.

19. Miller, R.K., et al., *In vitro patellofemoral joint force determined by a non-invasive technique.* Clin Biomech (Bristol, Avon), 1997. 12(1): p. 1-7.

20. Miller, R.K., et al., *In vitro measurement of patellofemoral force after three types of knee replacement.* J Bone Joint Surg Br, 1998. 80(5): p. 900-6.

21. Rudy, T.W., et al., *A combined robotic/universal force sensor approach to determine in situ forces of knee ligaments.* J Biomech, 1996. 29(10): p. 1357-60.

22. Li, G., et al., *Effect of combined axial compressive and anterior tibial loads on in situ forces in the anterior cruciate ligament: a porcine study.* J Orthop Res, 1998. 16(1): p. 122-7.

23. Fujie, H., et al., *Forces and moments in six-DOF at the human knee joint: mathematical description for control.* J Biomech, 1996. 29(12): p. 1577-85.

24. Fujie, H., et al., *The use of a universal force-moment sensor to determine in-situ forces in ligaments: a new methodology.* J Biomech Eng, 1995. 117(1): p. 1-7.

25. Fujie, H., et al., *The use of robotics technology to study human joint kinematics: a new methodology.* J Biomech Eng, 1993. 115(3): p. 211-7.

26. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

27. Hoher, J., et al., *In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study.* J Orthop Res, 1999. 17(5): p. 763-8.

28. Harner, C.D., et al., *Biomechanical analysis of a posterior cruciate ligament reconstruction. Deficiency of the posterolateral structures as a cause of graft failure.* Am J Sports Med, 2000. 28(1): p. 32-9.

29. Sakane, M., et al., *Relative contribution of the ACL, MCL, and bony contact to the anterior stability of the knee.* Knee Surg Sports Traumatol Arthrosc, 1999. 7(2): p. 93-7.

30. Sakane, M., et al., *In situ forces in the anterior cruciate ligament and its bundles in response to anterior tibial loads.* J Orthop Res, 1997. 15(2): p. 285-93.

31. Li, G., et al., *The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL.* J Biomech, 1999. 32(4): p. 395-400.

32. Fox, R.J., et al., *Determination of the in situ forces in the human posterior cruciate ligament using robotic technology. A cadaveric study.* Am J Sports Med, 1998. 26(3): p. 395-401.

33. Grood, E.S. and W.J. Suntay, *A joint coordinate system for the clinical description of three-dimensional motions: application to the knee.* J Biomech Eng, 1983. 105(2): p. 136-44.

34. Berger, R.A., et al., *Determining the rotational alignment of the femoral component in total knee arthroplasty using the epicondylar axis.* Clin Orthop, 1993(286): p. 40-7.

35. Berger, R.A., et al., *Malrotation causing patellofemoral complications after total knee arthroplasty.* Clin Orthop, 1998(356): p. 144-53.

36. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

Chapter 3

THE EFFECT OF AXIS CHOICE ON KNEE JOINT KINEMATICS

3.1 Introduction

Two flexion axes are widely used to describe knee joint motion: transepicondylar axis [1-4] and the geometric center axis [3]. The transepicondylar axis is defined as the axis connecting the most prominent points on the lateral and medial condyles while the geometric center axis is defined as the axis connecting the centers of the two femoral condyles. Geometric differences between these two axes have been noted in literature [3]; however, the variations in knee kinematics calculated using the two axes are unknown.

Recent cadaveric studies [5-7] examined the choice of flexion axes on knee rotation. One study compared the optimal flexion axis with the transepicondylar axis [7], where the optimal flexion axis was constructed by forming a line passing through the posterior femoral condyles. This study found that the transepicondylar axis closely approximates the optimal axis. However, another study examined the morphology of the distal femur and found that a common "cylindrical" axis (geometric center axis) represents a single, fixed, flexion/extension axis that is distinct from the transepicondylar axis [3]. Piazza and Cavanagh [8] examined the sensitivity of the screw-home knee motion due to errors caused by misaligning the rotational axis. They reported that misalignment of the joint coordinate system with respect to the corresponding rotation axis resulted in "kinematic crosstalk" thus greatly altering the knee kinematics measurements.

Even though different flexion axes have been defined and used in the literature, limited data is available that directly compares the kinematics data obtained by using the different flexion axes on the same knee motion. The objective of the study was to compare femoral translation and tibial rotation measured with respect to the transepicondylar axis (TEA) and the geometric center axis (GCA) from full extension to 150° of flexion. Three dimensional animated models of the human knee joint and a robotic testing system were utilized in this study.

3.2 Materials and Methods

3.2.1 Specimen Preparation

Six fresh frozen human cadaveric knee specimens (4 female knees, 2 male knees; average age 77 ± 7 years; age range 68-85) were used in this study. Prior to experimentation, radiographic images of each specimen were taken to assure good bone quality. Each specimen was manually flexed to assure full range of motion (0° to 150°). Specimens that did not have full range of motion were excluded from this study. Each knee joint specimen was approximately 50 cm long. The skin, subcutaneous tissue, joint capsule, ligaments and muscles were left intact. The fibula was secured to the tibia in its anatomical position. The specimen was hydrated regularly with 0.9% saline solution to avoid tissue dehydration. A more detailed description is given in section 2.3 above.

3.2.2 Kinematics Determination

Each specimen was manually pre-conditioned ten times prior to its installation on the robotic system. The passive path (i.e. no muscle activation, similar to manual surgeon controlled motion) for each specimen was determined by the robotic testing system at every 1° from full extension to 150° of flexion [4, 9-11]. After the passive path was determined, all soft tissue around the knee joint was removed, leaving the femur and tibia bones exposed. A visual inspection was performed to assure that the knee was healthy with no cartilage degeneration or soft tissue degradation.

After inspection, the geometry of the knee joint was determined in order to create a three dimensional model of the joint. The femoral medial and lateral collateral ligament insertion sites and the femoral and tibial surface contours were digitized (Microscribe® 3DX, Immersion, San Jose, CA); the digitization system has a spatial accuracy of 0.23 mm [11]. The digitized points and contours were imported into a three-dimensional (3D) modeling program (Rhinoceros®, Robert McNeel & Associates, Seattle, WA) to create a 3D model of the knee joint.

The TEA and GCA were determined using the complete 3D bone models. To construct the TEA (X_{TEA}), the center of the lateral collateral ligament was identified and the most prominence point within the insertion site was digitized. On the medial condyle, the prominence anterior to the sulcus point was selected and also digitized. The TEA was defined as a line connecting these two most prominent points [12, 13] (Figure 4A). The GCA (X_{GCA}) was constructed by fitting circles to the medial and lateral condyles and then connecting the centers of these circles with a line [3] (Figure 4A).

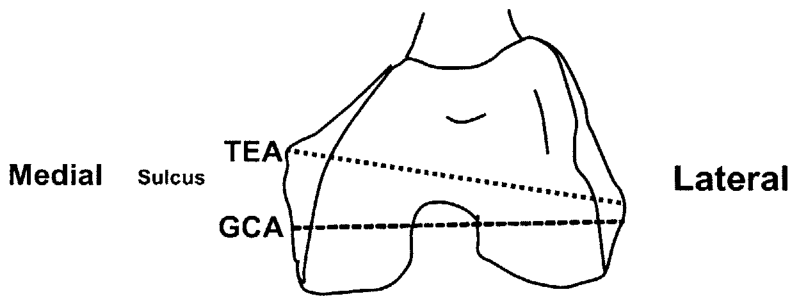


Figure 4A: The figure show an anterior view of the femur and the two axes used in this study to quantify knee kinematics. TEA ≡ transepicondylar axis; GCA ≡ geometric center.

In order to track the knee kinematics, three non co-linear points on both the femoral and tibial shafts were used to reference the position of the bones (knee kinematics) in three dimensional space. The reference points were digitized at every five degrees from full extension to 150° of flexion by replaying the recorded passive path of the knee on the robotic testing system. These digitized reference points were placed on the 3D bone models. The insertion sites and surface contour data points were related to the tibial and femoral shaft reference points in order to relate the bony geometry to the knee kinematics, respectively, along the flexion path.

In each experiment, the longitudinal axis of the tibia was selected to represent the tibial mechanical axis such that the rotation about this axis corresponds to internal-external motion. Throughout this experiment, the femur and tibia were considered rigid bodies, thus the axes remain fixed on the bones. Using these 3D knee models, the anterior-posterior femoral translation was measured using the TEA and GCA with the knee position at full extension as a reference [4, 9-11, 14]. Internal-external tibial rotation was calculated using the convention as defined by Grood and Santay [15] (Figure 4B).

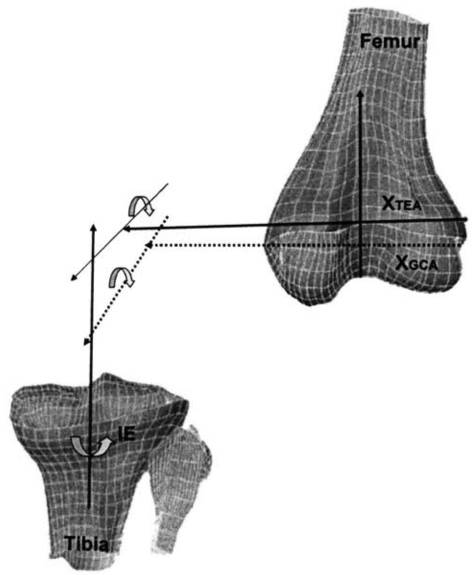


Figure 4B: Axis definition: Two floating axes one based on the TEA and the other based on the GCA.

3.2.3 Statistical Analysis

Two-way repeated measures Analysis of Variance (ANOVA) were used to detect whether axis choice (TEA and GCA) had a significant affect on femoral translation and tibial rotation. Knee position at full extension was used as the reference. Student-Newman-Keuls post-hoc tests were done to detect any statistical differences between the groups. Differences were taken to be significance for p-values less than 0.05.

3.3 Results

3.3.1 Axis Differences

Comparing the TEA and the GCA, it was found that the distance between the axes on the medial condyle to be farther apart than on the lateral condyle (Figure 5). The mean distance between axes on the medial condyle was 15.5 ± 2.2 mm and between the axes on the lateral condyle was 6.9 ± 0.9 mm. The two axes formed an angle of 4.0 ± 0.8°.

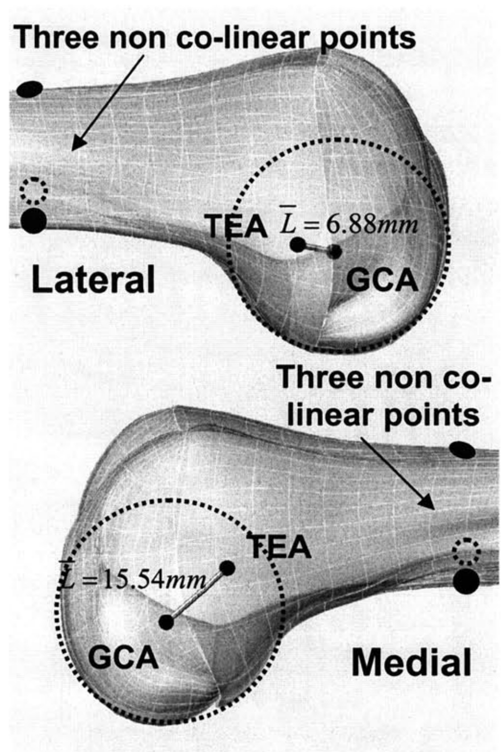


Figure 5: Axis differences: When comparing the axes anatomically, a large difference was noted between the positioning of the two axes on the medial side when compared to the positioning of the two axes on the lateral side. The small dotted circles represent the points on the far side of the bones.

3.3.2 Femoral Translation

The mean medial GCA maintained anterior translation from full extension to 120° and posterior translation from 120° to 150° (Figure 6). The peak anterior translation for the GCA was approximately at 50° (-7.5 ± 4.3mm). At 90°, the medial end of the GCA was situated -2.5 ± 9.0mm anteriorly while at 150°, the GCA medial end was positioned 5.4 ± 5.6mm posteriorly. The mean medial TEA values demonstrated minimal anterior-posterior translation from 0° to 50° (3.6 ± 4.4mm) then steady posterior translation from 50° to 150°. Using the TEA, the medial end was positioned 19.0 ± 10.0mm and 32.7 ± 9.1mm at 90° and 150° of knee flexion, respectively. There was a statistical significant difference ($p<0.05$) between the GCA and the TEA translation on the medial end for all flexion angles expect at full extension.

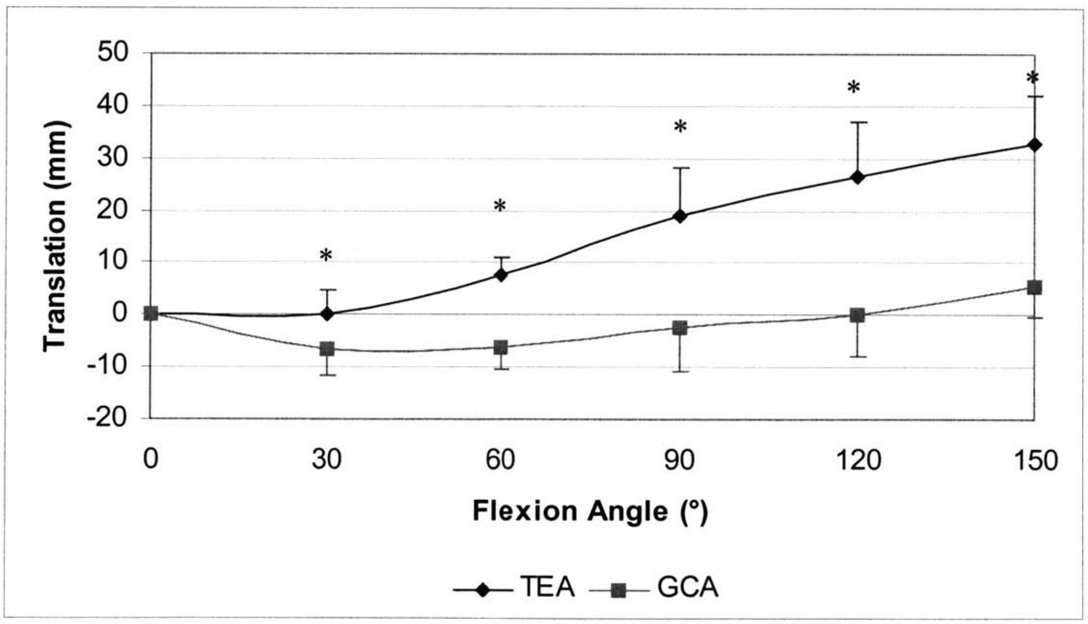


Figure 6: Graphic representation of mean transepicondylar axis (TEA) vs. mean geometric center axis (GCA) medial translations. The medial condyle of the TEA maintained a greater (+) posterior femoral translation throughout the passive path when compared to the medial condyle of the GCA. Error bars represent one standard deviation from the mean (* $p<0.05$).

The mean values for the lateral ends for both axes maintained consistent posterior translation throughout flexion (Figure 7). At 90°, the lateral end of the GCA was placed 15.6 ± 8.1mm posteriorly compared to full extension and continued to move posteriorly with flexion, peaking at 150° (31.7 ± 6.9mm). Using the TEA, the lateral end was located 25.0 ± 8.5mm and 42.1 ± 8.2mm posteriorly at 90° and 150°, respectively, compared to full extension. The translation of the lateral end of the GCA and TEA was statistically different ($p<0.05$) throughout the knee flexion range, expect at full extension.

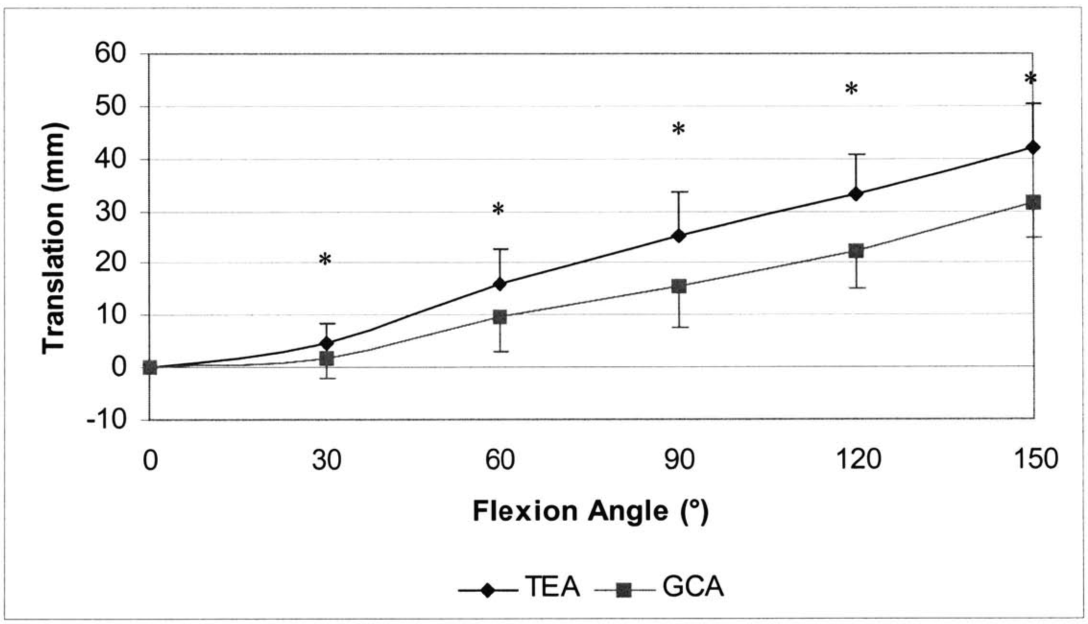


Figure 7: Graphic representation of mean transepicondylar axis (TEA) vs. mean geometric center axis (GCA) lateral translations. The lateral condyle of the TEA maintained a greater (+) posterior femoral translation throughout the passive path when compared to the medial condyle of the GCA. Error bars represent one standard deviation from the mean (* $p<0.05$).

3.3.3 Tibial Rotation

The mean GCA and TEA tibial rotation values followed similar trends with the mean GCA tibial rotation maintaining the larger value throughout the entire range of knee flexion (Figure 8). At 60°, the mean internal tibial rotation using the GCA was 12.0° ± 7.4° while using the TEA, the internal tibial rotation was 6.5° ± 7.2°. Using the

GCA and TEA, the tibia internally rotated by 13.8° ± 10.2° and 4.8° ± 9.4°, respectively, at 90°. Internal tibial rotation continued to rise reaching a peak value of 19.9° ± 6.9° (GCA) and 7.2° ± 5.7° (TEA) at 150° of flexion. Except at full extension, the tibial rotation observed from the GCA was significantly larger ($p<0.05$) than that observed using the TEA.

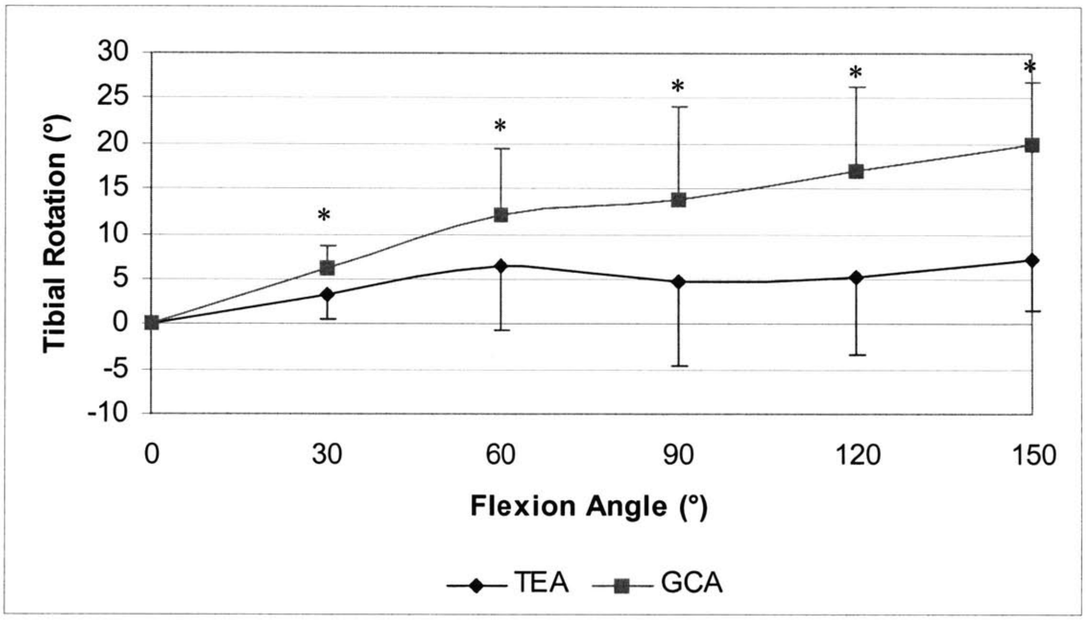


Figure 8: Graphic representation of mean transepicondylar axis (TEA) vs. mean geometric center axis (GCA) tibial rotations. The GCA maintained a greater degree of internal (+) tibial rotation compared to the TEA throughout passive flexion. Error bars represent one standard deviation from the mean (* $p<0.05$).

3.4 Discussion

Numerous investigations have presented various knee kinematics data. For example, several researchers [16, 17] have reported that with increasing of flexion, the lateral condyle translates posteriorly while the medial condyle shows minimal translation. They concluded that the tibia internally rotated with increasing flexion. Todo et al [18] reported that although femoral rollback did occur in the unloaded normal knee, its magnitude was small (~2mm). Most et al [9] reported that on the passive path, the intact knee internally rotated with increasing flexion angle reaching a maximum of 10.4° ± 9.5° at 120° flexion. Asano et al [19] found that with increasing knee flexion, external femoral rotation (equivalent to internal tibial rotation) increased. They reported that at 120°, the femur externally rotated by 23.8° ± 4.8° [19]. Aside from the different loading conditions used in these studies, the different axes (coordinate systems) may cause systematic differences with the reported knee motion values. Piazza and Cavanagh [8] has demonstrated that the screw-home mechanism of the knee could be noticeably affected by the axes of the knee joint.

This current study examined the sensitivity of kinematics data to two widely used flexion axes TEA and GCA. The TEA has been used in total knee arthroplasty for component alignment [12, 13] and knee kinematics studies [4, 9, 10, 14] due to identifiable bony landmarks. Many researchers have also reported kinematics data using GCA [16, 20-23] as it represents the posterior geometry of the femoral condyle [3]. Recently, several studies have examined the characteristics of these flexion axes [3, 23]. Eckhoff et al [3] has reported a distinct difference in these two axes. The data presented in our study also noticed that these two axes were not parallel but rather formed an angle of 4.0° ± 0.8° (Figure 4A). Though the geometry was compared, the effect of geometry on knee kinematics was not compared.

This study compared the femoral condyle translation and tibial rotation of the knee using the TEA and GCA from full extension to 150° of flexion. The results of the lateral condyle for both axes concurred with the results published by Hollister [6]. The plots of the lateral end of the axes produced approximate concentric arcs, indicating that the femur translated posteriorly with increasing knee flexion (Figure 9). However, the medial end of the TEA showed monotonic posterior translation for flexion angles between 30° and 150° (Figure 9). The medial end of the GCA first translated anteriorly at low flexion angles and peaked at 50° of flexion. It then proceeded to translate posteriorly beyond 50° (Figure 9). In both flexion axes, the lateral condyle translation was greater than the medial condyle translation throughout the entire range of flexion, resulting in internal tibial rotation with increasing flexion.

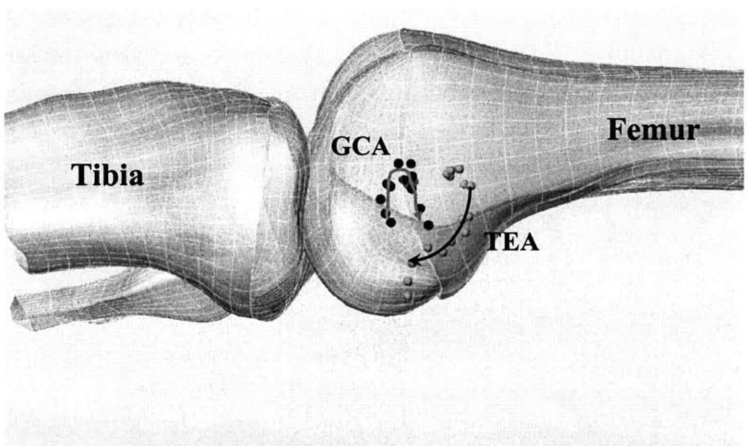


Figure 9: Graphic representation of femoral translation on the medial side using the TEA and the GCA. The TEA showed monatomic posterior translation while the GCA first translated anteriorly (<50°) and then proceeded to translate posteriorly.

The difference in the kinematics is directly related to the location of the medial and lateral points that were used in constructing the TEA and GCA axes. On the lateral side, the TEA and GCA ends are close to each other (approximately 7 mm apart); therefore, the lateral condyle translation is similar for both axes. However, on the medial side, a significant difference between the location of the two axes exists (approximately 15 mm apart); therefore, the medial condyle translation was different for both axes. Tibial rotation obtained from the two axes was also different. It was found that the tibial rotation using the GCA was greater than that recorded using the TEA throughout the entire range of flexion. As the medial femoral end of the GCA remain almost stationary through the entire range of flexion when compared to the medial end of the TEA, the lateral femoral end of the GCA translated posteriorly at an equivalent rate as the lateral end of the TEA. Consequently, the differential motion between the medial and the lateral sides measured using the GCA was greater than that measured using the TEA; yielding a greater internal tibial rotation for the GCA.

In conclusion, the effect of TEA and GCA on knee kinematics was examined in this study. The results suggest that the kinematics calculation is sensitive to the selection of flexion axis. While the TEA may result in higher posterior translation in medial femoral condyle, the GCA can lead to a higher tibial rotation. Both methods are adequate to describe knee motion as long as a clear definition of the flexion axis is given when reporting knee joint kinematics.

3.5 References

1. Yoshino, N., et al., *Computed tomography measurement of the surgical and clinical transepicondylar axis of the distal femur in osteoarthritic knees.* J Arthroplasty, 2001. 16(4): p. 493-7.
2. Olcott, C.W. and R.D. Scott, *The Ranawat Award. Femoral component rotation during total knee arthroplasty.* Clin Orthop, 1999(367): p. 39-42.
3. Eckhoff, D.G., et al., *Three-dimensional morphology of the distal part of the femur viewed in virtual reality.* J Bone Joint Surg Am, 2001. 83-A(Suppl 2(Pt 1)): p. 43-50.
4. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.
5. Feikes, J., D. Wilson, and J. O'Conner. *The Unique Track of Intact Passive Knee Motion as a Kinematic Baseline.* in *Orthopaedic Research Society.* 1998. New Orleans, Louisiana.
6. Hollister, A.M., et al., *The axes of rotation of the knee.* Clin Orthop, 1993(290): p. 259-68.
7. Churchill, D.L., et al., *The transepicondylar axis approximates the optimal flexion axis of the knee.* Clin Orthop, 1998(356): p. 111-8.
8. Piazza, S.J. and P.R. Cavanagh, *Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment.* J Biomech, 2000. 33(8): p. 1029-34.

9. Most, E., et al., *Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty.* Clin Orthop, 2003(410): p. 101-13.

10. Most, E., et al., *The Kinematics of Fixed and Mobile Bearing Posterior Stabilized Total Knee Arthroplasty.* Clin Ortho, 2003. 416: p. 1-11.

11. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

12. Berger, R.A., et al., *Determining the rotational alignment of the femoral component in total knee arthroplasty using the epicondylar axis.* Clin Orthop, 1993(286): p. 40-7.

13. Berger, R.A., et al., *Malrotation causing patellofemoral complications after total knee arthroplasty.* Clin Orthop, 1998(356): p. 144-53.

14. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

15. Grood, E.S. and W.J. Suntay, *A joint coordinate system for the clinical description of three-dimensional motions: application to the knee.* J Biomech Eng, 1983. 105(2): p. 136-44.

16. Iwaki, H., V. Pinskerova, and M.A. Freeman, *Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee.* J Bone Joint Surg Br, 2000. 82(8): p. 1189-95.

17. Nakagawa, S., et al., *Tibiofemoral movement 3: full flexion in the living knee studied by MRI.* J Bone Joint Surg Br, 2000. 82(8): p. 1199-200.

18. Todo, S., et al., *Anteroposterior and rotational movement of femur during knee flexion.* Clin Orthop, 1999(362): p. 162-70.

19. Asano, T., et al., *In vivo three-dimensional knee kinematics using a biplanar image- matching technique.* Clin Orthop, 2001(388): p. 157-66.

20. Blankevoort, L., R. Huiskes, and A. de Lange, *The envelope of passive knee joint motion.* J Biomech, 1988. 21(9): p. 705-20.

21. Blankevoort, L., R. Huiskes, and A. de Lange, *Helical axes of passive knee joint motions.* J Biomech, 1990. 23(12): p. 1219-29.

22. Hill, P.F., et al., *Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI.* J Bone Joint Surg Br, 2000. 82(8): p. 1196-8.

23. Freeman, M.A. and V. Pinskerova, *The movement of the knee studied by magnetic resonance imaging.* Clin Orthop, 2003(410): p. 35-43.

Chapter 4

INTACT KNEE BIOMECHANICS

4.1 Motivations and Objectives

Total knee arthroplasty (TKA) has been a common and, in many cases, a successful surgical procedure to alleviate patient's joint pain. Moreover, the surgery aims at allowing patients to return to their activities of daily living by restoring their normal (pre-operative) range of motion. However, review of the literature reveals that patients rarely flex beyond 120° after contemporary TKA [1-12]. To better understand the limitation of current TKA procedure imposes on knee motion as compared to the pre-operative range of motion, it is important to take a step back, and appreciate the nature of the intact (unreconstructed) knee, particularly at high knee flexion.

The motion of the intact knee is guided by the geometry of the articulating surfaces and the soft tissue about and within the knee joint (including the ligaments, posterior soft tissue, and the menisci) [11, 12]. It is reported that the human intact knee is capable of flexion up to 160° [13-15]. The presence of posterior osteophytes (bony ridges or spurs) and tight extension mechanism (quadriceps contracture) has been shown to limit knee flexion. However, limited quantitative data concerning the behavior of the intact knee beyond 120° has been reported in the literature. This chapter contains a collection of experimental data which examines the nature of the intact knee. The data presented here address the following questions:

(1) Do femoral translation and tibial rotation continue after 120° of knee flexion?

(2) What guides knee motion beyond 120° of flexion?

(3) What is the role of the soft tissue around the knee joint (cruciate ligaments, menisci, posterior soft tissue, muscles, skin, and fat)?

At the outset, our hypotheses were:

(1) The kinematics of the knee is closely restored after TKA; yet, the articular contact is altered after the surgery.

(2) The posterior cruciate ligament (PCL) promotes posterior femoral translation beyond 120° of flexion.

(3) The quadriceps and hamstrings muscles guide the knee at deep flexion.

This investigation makes use of human cadaveric specimen in combination with a robotic testing system [16].

4.2 Literature Review

Considerable number of papers have been published describing the kinematics of the intact knee in the flexion range from 0° to 120° [17-37]. In a fluoroscopic analysis of healthy subjects, Komistek et al [25] examined the medial and lateral femoral condyles contact translation under weight bearing as a function of knee flexion. They reported that the lateral condyle translates more posteriorly than the medial condyle with increasing knee flexion, signifying the presence of internal tibial rotation. Similar results were reported by other researchers [29, 30, 35]. Dennis et al [36] investigated tibiofemoral contact in the sagittal plane using fluoroscopic videos, image matching and discrete digitization of healthy subjects. They reported that at full extension, the femur contacted the tibia anterior to the midpoint in the sagittal plane (6.49 mm) and then translated posteriorly with increasing flexion reaching a value of -7.68 mm at 90° of flexion (total translation of 14.17 mm). Todo et al [17] analyzed the anteroposterior and rotational movement of the femur as a function of knee flexion in ten healthy male subjects. In contrast to the previous studies, they found that an unloaded femur rolled back but its magnitude was minimal (2.4 mm on the medial side and 5.4 mm on the lateral side at 90°). Piazza and Cavanagh [38] measured the screw-home motion of the knee (defined as external rotation of the tibia with respect to the femur during knee extension) and its sensitivity to errors in misaligning the rotational axis. They reported that "kinematic crosstalk" greatly alter the measurements of knee kinematics as a results of misaligned joint coordinate system with the corresponding rotation axis.

Limited data regarding the behavior of the intact knee beyond 120° is available [14, 15, 34, 39, 40]. Hefzy et al [39] measured the kinematics (contact) of the intact knee using bi-planar radiographs. They found that an asymmetric rolling motion occurs during kneeling and that the tibia rotated internally. Furthermore, they observed that at high knee flexion, the tibiofemoral contact occurs in the most proximal aspect of the posterior condyles. Nakagawa et al [15] used a Magnetic Resonance Imaging (MRI) technique to examine the kinematics of 20 healthy adult volunteers. They concluded that the lateral condyle translated more posteriorly than the medial condyle with increasing knee flexion and at 162°, the lateral condyle lost its normal contact with the tibia. Nonetheless, attempts to better understand the six degrees-of-freedom kinematics of the knee at high flexion (>120°) remain of great challenges among researchers.

The biomechanical function of the PCL in the intact knee between 0° and 120° has been examined by many research groups [31, 41-47]. The roles of the PCL, to name few, include resisting posterior tibial translation beyond 30°, providing knee stability primarily at mid-flexion range, and promote femoral rollback. It has been shown that under combined muscle loads, the PCL carries minimal force at full extension and at 120° and peaks at approximately 90° of knee flexion [31, 45, 48]. Below 120° of knee flexion, partial or complete tear in the PCL was shown to significantly decrease the amount of femoral translation under external loads [31, 33, 44]. Nagura et al [14] examined the mechanical loads at the knee joint of 19 subjects during deep flexion. They concluded that "*the posterior cruciate ligament should have a substantial role during deep flexion, since there was a large posterior load that must be sustained at the knee*" [14]. However, little is know on the biomechanical function of the PCL and other soft tissue beyond 120°.

In recent years, a robotic testing system was used to study intact knee kinematics in multiple degrees of freedom and to measure the in situ forces in the knee ligaments [18, 19, 45, 48, 49]. However, most studies were limited to the flexion range of 0° to 90° and did not include information on knee replacements. In this dissertation, the concept of the robotic system was expanded to allow wider range of flexion and multiple knee reconstructions on the same specimen thereby limiting intra-specimen variation.

4.3 Testing Protocol

4.3.1 Experimental Set-Up

Depending on a given study, between eight and fifteen fresh frozen cadaveric knee specimens were used. Prior to testing, each specimen was thawed overnight at room temperature. To assure bone quality and absence of all disease, each specimen was radiographed in both the anterior-posterior (AP) and lateral views. Each specimen was manually flexed to verify full range of flexion (0° - 150°). Specimens that did not have full range of motion were excluded from this study. The fibula was fixed to the tibia in its anatomical position by a cortical bone screw. The femur and tibia were cut to approximately 25 cm in length from the joint line. The ends were stripped of musculature, and potted in bone cement. The remaining soft tissues surrounding the joint (skin, knee ligaments, joint capsule, and musculature) were otherwise left intact. The knee was manually flexed ten times between full extension and full flexion. To prevent tissue hydration, the specimen was hydrated regularly with 0.9% saline.

The specimen was secured in thick-walled aluminum cylinders and then mounted on a robotic testing system (Figure 10). The femoral cylinder was mounted and rigidly fixed in a specially designed clamp that enables 6 degrees of freedom (DOF) positioning relative to the base of the manipulator. The tibial cylinder was then mounted to the 6 DOF load cell, which in turn was rigidly secured to the end-effector of the robotic manipulator (Figure 10). In this arrangement, the tibia was moved by the manipulator to allow 6 DOF knee motion. The robotic manipulator can determine the complex motion of the knee specimen in response to external loads [16, 49]. The knee was aligned so that the load cell measured three forces and three moments along and about a Cartesian coordinate system defined by the longitudinal axis of the tibia (*x*), the transepicondylar (medial-lateral) axis of the femur (*y*), and the anterior-posterior axis of the knee (*z*). The midpoint of the transepicondylar line represented the origin of the coordinate system (Figure 11).

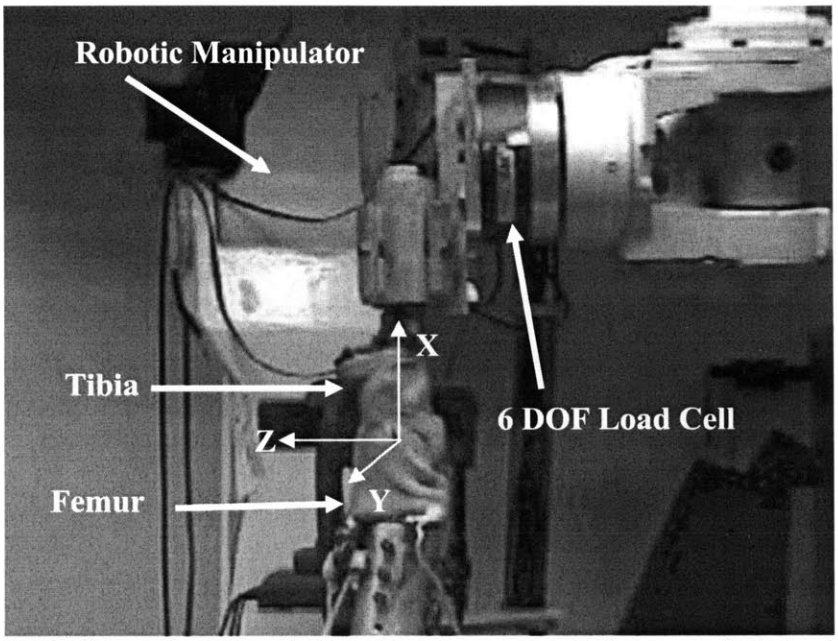


Figure 10: The figure shows the robotic testing system set up. The system includes the robotic manipulator, 6 degrees-of-freedom load cell, and the intact knee.

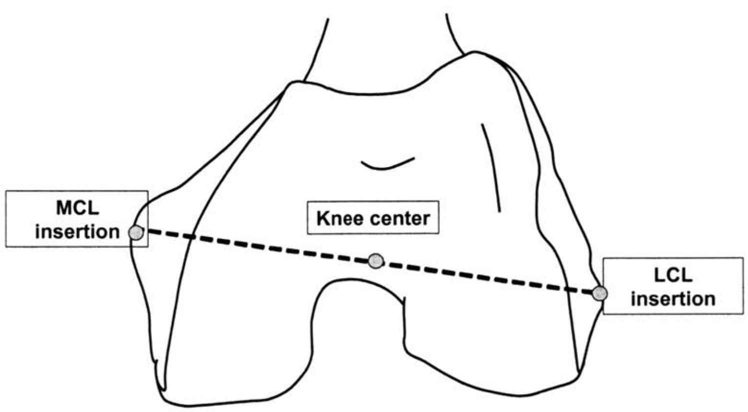


Figure 11: The figure outlines the transepicondylar axis which is the line connecting the insertion site of the medial and lateral collateral ligaments (LCL≡ lateral collateral ligament; MCL≡ medial collateral ligament).

At full extension, this coordinate system was used to describe the position of both the femur and tibia. The overall testing protocol is described in Figure 12.

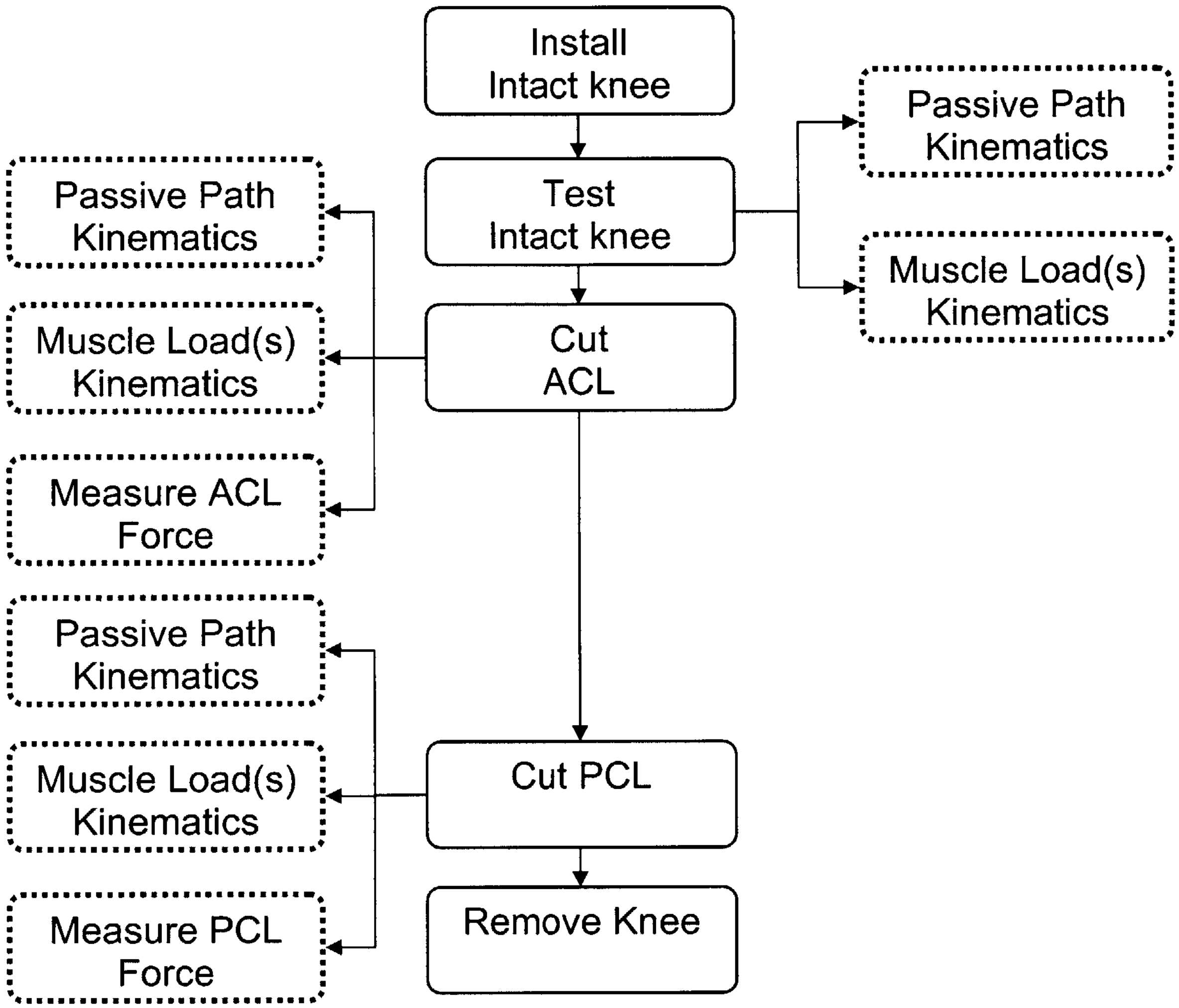


Figure 12: Overall testing protocol for testing the biomechanics of intact knees. ACL ≡ anterior cruciate ligament; PCL ≡ posterior cruciate ligament.

4.3.2 Determination of the Passive Path

The initial position of the knee at a selected flexion angle is crucial for measuring the joint kinematics in response to external loads. The passive position of the knee was used as the initial position for the application of external loads. At a selected flexion angle, a passive position was defined as the position of the knee where it carried minimal load with no external forces applied. A series of passive positions was defined during passive knee flexion/extension between full extension and 150°. This was done using the force-moment control of the test system at one-degree increments of knee flexion. Knee positions in the remaining five degrees-of-freedom were determined such that the residual forces and moments on the intact knee were minimal (below 5 N and 0.5 N-m, respectively). This series of positions form a passive path, which was repeated. The path was then used as the reference positions for the application of simulated muscle loads during subsequent tests. In all studies, the knee testing was limited to 150° of flexion so as to avoid impingement of the fixtures at high flexion angles.

4.3.3 The Kinematics of the Knee under Various Muscle Loads

After the passive path was determined, muscle loads were applied to the knee at the predetermined flexion angles of 0°, 30°, 60°, 90°, 120°, and 150°. Simulation of isolated quadriceps and hamstring muscle function and their co-contraction was performed as previously described by Li et al. [19]. Nylon ropes were sutured to the quadriceps, semitendinosus, semimembranosus and biceps femoris tendons and weights were hung from these ropes through a system of pulleys. Three different muscle loads were simulated: 1) an isolated quadriceps load of 400N; 2) a combined quadriceps/hamstring load of 400N/200N; and 3) an isolated hamstring load of 200N (that is, 100N for the medial hamstrings, and 100N for the lateral ones). The motion of the tibia in response to these muscle loads in the remaining five degrees-of-freedom was measured. Thus, the kinematic response (femoral translation and tibial rotation) of the knee joint to muscle loading was recorded at each flexion angle [19].

4.3.4 The Contribution of the Cruciate Ligaments

The anterior cruciate ligament (ACL) was transected via a small medial arthrotomy at 30° of knee flexion. Careful attention was paid to avoid damage to other structures. After cutting the ACL, the arthrotomy and skin were closed in layers. The kinematics of the intact knee under the simulated muscle loads were replayed on the same specimen at each of the selected flexion angles. The force transmitted through the knee joint was measured by the load cell. The difference between this force and the force measured in the intact knee represented the in-situ force in the ACL using the principle of superposition [19].

Next, with the knee in 90° of flexion, the PCL was transected via a small medial arthrotomy, carefully avoiding damage to other structures. After cutting the PCL, the arthrotomy and skin were closed in layers. The same kinematics of the intact knee under the three external loads were replayed on the same specimen at each of the selected flexion angles. The forces transmitted through the knee joint were measured by the load cell. The difference between this force and the force measured in the intact knee represented the in situ force in the PCL, using the principal of superposition [19].

4.4 Results

4.4.1 Intact Knee Kinematics on the Passive Path

On the passive path, the femur translated posteriorly with increasing knee flexion reaching an observed peak value at full flexion (Figure 13). Flexion angle was found to have a statistical significant affect on the amount of femoral translation. For example, at

30° of knee flexion, the intact knee center translated posteriorly by 1.1 ± 2.2 mm (mean ± SD) which was statistically less than the translation observed at 60° (6.9 ± 4.5 mm). At 120°, the femur continues to translate posteriorly reaching a value of 20.9 ± 7.5 mm. Peak value was observed at 150° (29.6 ± 9.2 mm), which was statistically larger than at lower flexion angles.

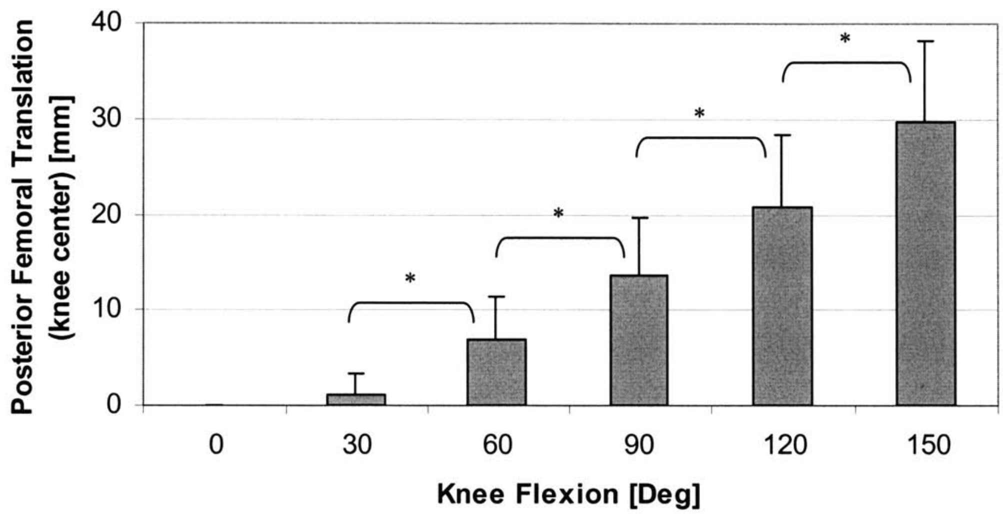


Figure 13: Posterior femoral translation as a function of knee flexion on the passive path. * denotes a p value less than 0.05.

Direct correlation between internal tibial rotation and flexion angle was observed (Figure 14). Minimal tibial rotation was observed for flexion angles less than 30° (<5 degrees). At 90°, the tibia internally rotated by 7.2° ± 6.5° which was statistically more significant (p=0.023) than the rotation observed at 60° (4.0° ± 4.4°). With further increase in knee flexion, the tibia continued to internally rotate reaching a value of 11.5° ± 7.9° at 150° of knee flexion.

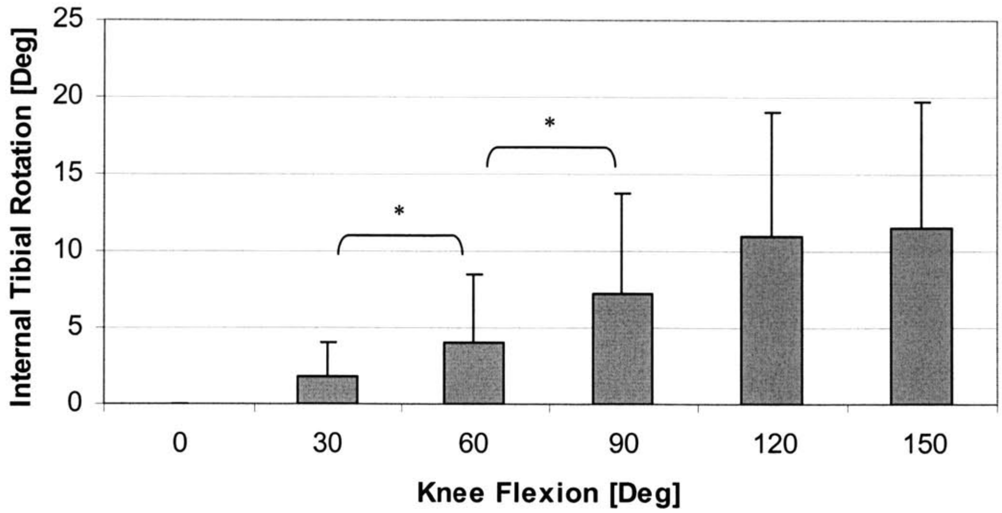


Figure 14: Internal tibial rotation as a function of knee flexion on the passive path. * denotes a p value less than 0.05.

4.4.2 Intact Knee Kinematics under Combined Muscle Load

It has been suggested by many researchers [15, 32, 33, 50, 51] that as the knee flexes, a combined translation and rotation motion can be observed. To capture this combined motion, the translation of the lateral and medial femoral condyles were tracked and compared. Figure 15 plots the average and standard deviation values of posterior femoral translation as a function of knee flexion for the lateral and medial femoral condyles. At full extension, the lateral condyle was positioned 3.3 ± 1.6 mm posteriorly when compared to unloaded intact knee at full extension. With increasing flexion, the lateral condyle gradually translated posteriorly, reaching a value of 13.8 ± 7.0 mm at 90°. Beyond 90°, posterior translation continued, reaching a maximum value of 31.9 ± 12.5 mm at 150°. For flexion angles less than 90°, the medial condyle hardly moved posteriorly and in fact, simply acted as a pivot point for the knee. At 90°, the posterior femoral translation recorded for the medial condyle was 9.1 ± 6.8 mm. With further knee flexion, the medial condyle progressively translated posteriorly reaching a peak value of 22.9 ± 11.3mm at 150°.

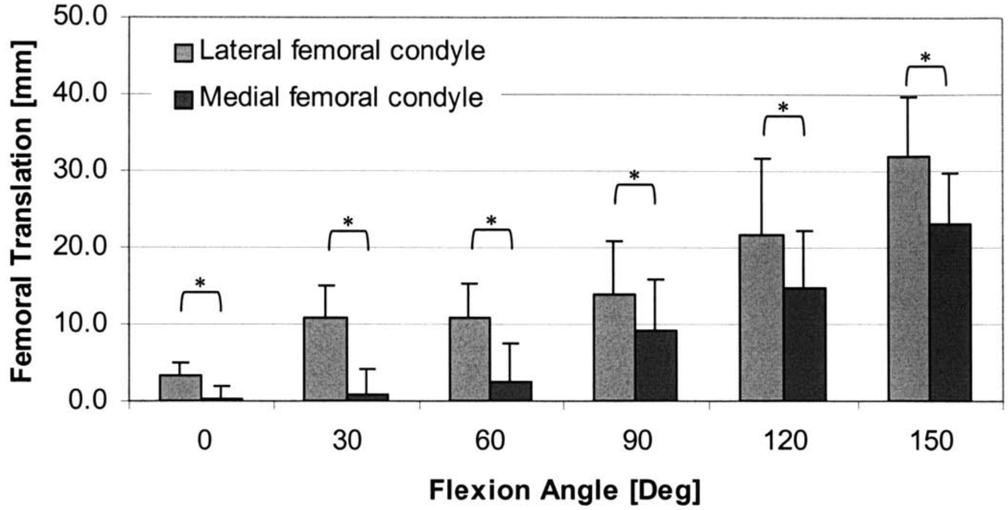


Figure 15: The figure shows the posterior femoral translation (+) of the lateral and medial femoral condyles as a function of knee flexion. * denotes a p value less than 0.05.

Throughout the entire range of flexion, the lateral femoral condyle translated significantly more posteriorly ($p<0.05$) compared to the medial condyle. This differential motion between the lateral condyle and the medial condyle signifies the presence of internal tibial rotation. Figure 16 depicts this differential motion superimposed on a cross section of the tibia. It was observed that most of the internal tibial rotation occurred at low flexion angles (<90). For example, at 30° of flexion, the tibia rotated 11.0° ± 5.7° when compared to the unloaded knee at full extension. Beyond 90°, tibial rotation remained relatively constant.

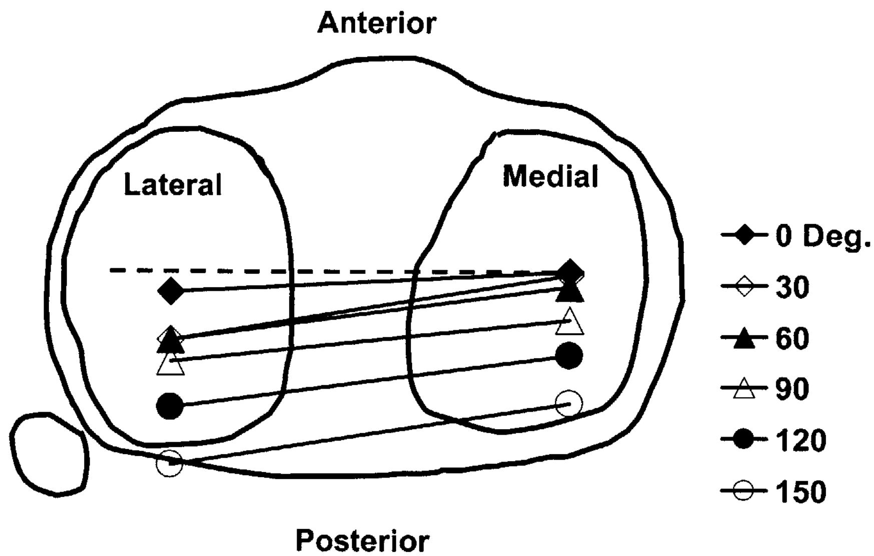


Figure 16: The figure illustrates the differential posterior femoral translation motion of lateral and medial condyles on the cross section of the tibia. The higher posterior translation of the lateral condyle compared to the medial condyle indicates internal tibial rotation.

4.4.3 The Role of the ACL

The in-situ forces in the ACL were highest at early flexion and then decreased rapidly with increasing knee flexion (Figure 17). Under the quadriceps load, the in-situ ACL force was 63.9±33.4 N at full knee extension. This increased to a maximum of 71.7±27.9 N at 30° and then dropped off rapidly to 25.1±17.0 N at 60° and to 11.7±3.0 N at 90°. From that point, the ACL forces slowly increased with increasing flexion and reached 29.6±12.9 N at 150° which was significantly higher than the forces at 90° and 120°.

Under the combined muscle load, ACL forces were maximal at full extension and 30° of flexion at 52.3±24.4 N and 46.9±19.2 N, respectively (Figure 17). Like the forces under the isolated quadriceps load, they again decreased rapidly at 60° and did not significantly change throughout the remaining range of flexion, even though the mean value increased slightly after 90°.

Under an isolated hamstrings load, the in-situ forces in the ACL reached a peak value of 32.3±20.9 N at 30° (Figure 17). After 60° of flexion, there were no statistically significant changes in the forces of the ACL as the flexion angle increased.

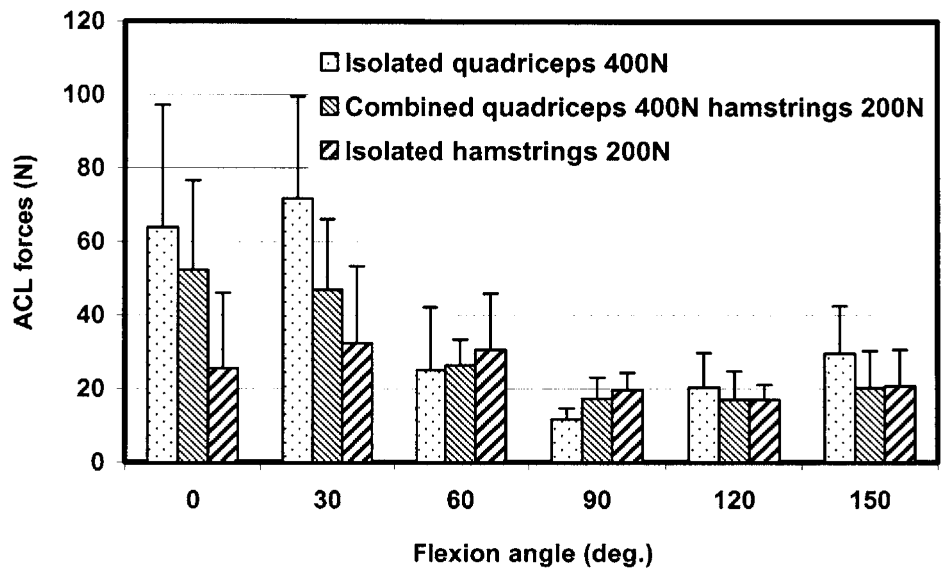


Figure 17: In-situ forces of the ACL under various muscle loads.

4.4.4 The Role of the PCL

The in-situ forces in the PCL changed with flexion of the knee and the applied muscle loads (Figure 18). Under the isolated quadriceps load, the PCL forces measured about 20 N throughout the entire range of flexion.

Under the combined muscle loads, the PCL force was similar to the force under the isolated quadriceps load at full extension (24.4±16.6 N). However, there was a significant rise in force with every flexion angle to a peak of 73.2±28.7 N at 90° followed by a significant decrease to 31.9±14.2 N at 150° of flexion (Figure 18).

The isolated hamstrings load resulted in a similar in-situ force in the PCL at full extension compared to all other muscle loading conditions (Figure 18). There was a significant increase, however, in the magnitude of the force from 0° to 30°, and from 30° to 60°, peaking at 74.8±27.9 N at 90° of flexion. The force then decreased steadily reaching 36.8±17.8 N at 150° of flexion.

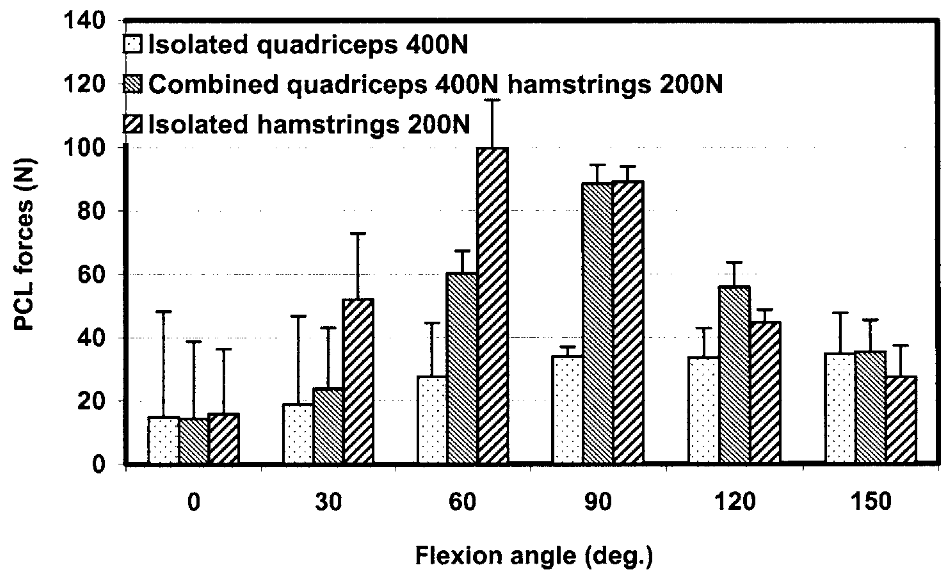


Figure 18: In-situ forces of the PCL under various muscle loads.

4.4.5 The Role of the Soft Tissue

The force in the PCL was minimal at low flexion angles, peaked at 90°, and diminished beyond 90°. The PCL force at full extension (0°) and full flexion (150°) were similar. This signifies the importance of the PCL in the mid flexion range but not necessarily in high flexion angles. The force in the ACL was maximal at low flexion angles (<30°) and decreased with further increase in knee flexion. This signifies the importance of the ACL in the low flexion range but, as with the PCL, not necessarily in high flexion angles. The forces in both cruciate ligaments at high knee flexion (150°) were minimal on the passive path and under the various muscle loads; yet, posterior femoral translation continued to occur at high flexion angles. It has been proposed that the posterior soft tissue (posterior capsule, menisci, hamstring muscles, skin, and fat) guide the knee beyond 120°.

When replaying the intact knee kinematics we observed that beyond 120°, the posterior soft tissue was compressed between the femoral shaft and the posterior edge of the tibia. We believe that this posterior soft tissue compression plays an important role in promoting further posterior femoral translation in high knee flexion. This becomes possible due to the fact that the compressed posterior soft tissue pushes the tibia anteriorly (equivalent to posterior translation of the femur).

The contribution of the menisci to knee motion and stability was examined by dissecting all the soft tissue around the knee joint, leaving only the menisci intact. This was done while the specimen was still mounted on the robotic testing system. The pre-recorded kinematics was replayed and we observed the motion of the menisci from full extension to 150° of flexion. The medial meniscus remained attached to the tibial plateau throughout the entire range of knee motion, providing an "extended" surface on the tibia that allows the femur to rollback over the tibia in a stable manner at high knee flexion (Figure 19). However, on the lateral side, between full extension and 120°, the lateral meniscus followed the tibia closely, but beyond 120°, the meniscus elongated posteriorly on the tibial plateau, extending beyond the tibial surface to provide stability for the knee (Figure 19).

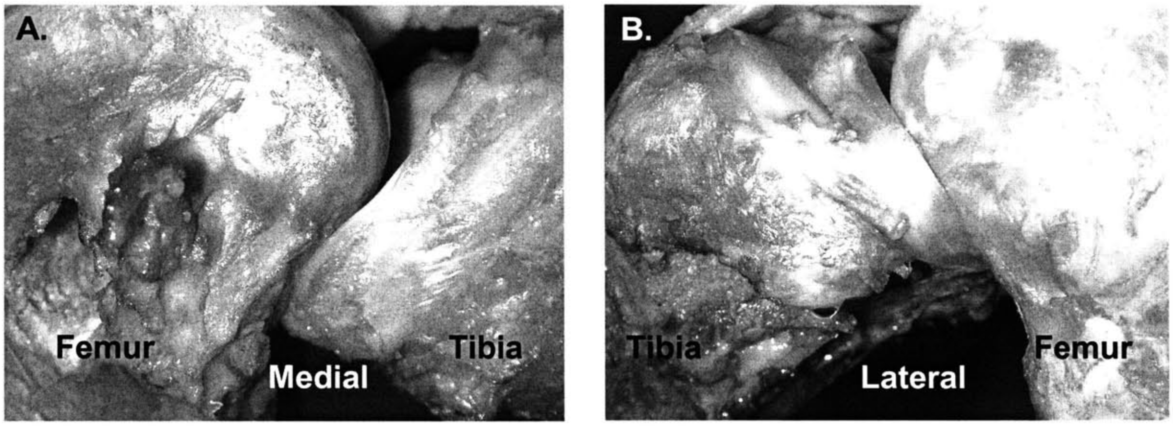


Figure 19: The position of the A) medial and B) lateral meniscus at high knee flexion (150°).

4.5 Conclusions

The kinematics of the intact knee [17, 19, 21-24, 26-28, 33, 52] and the forces in the cruciate ligaments (ACL [19, 53-55] and PCL [31, 41, 42, 44-46]) between full extension and 120° of flexion have been investigated in the literature. Until now, little was known about the kinematics, the kinetics, and the behavior of the intact knee beyond 120°. This study is the first to quantify these intact knee variables at high knee flexion (>120°).

The lateral and medial condyles of the intact knee translated posteriorly with increasing knee flexion, reaching an observed peak value at full flexion (150°). Throughout the entire range of flexion, the lateral condyle translation exceeded the posterior translation observed in the medial condyle signifying the presence of internal tibial rotation. The data in this study showed that most of the knee rotation is achieved by the first 30° of knee flexion. Rotation remained relatively constant beyond 30° of flexion with slightly elevated internal rotation at high knee flexion. We attribute this knee behavior to the fact that at 30°, the knee is least constrained and that the effect of muscle load application is most noteworthy.

The results from this study show that both cruciate ligaments (ACL and PCL) played a minimal role in restraining knee motion in high flexion angles. The ACL showed to carry peak load around 30° of flexion, which is consistence with other published data [19]. The PCL showed to carry highest load in the mid flexion range (90°), which is also consistence with other published data [45, 56].

The effect of muscle load on knee kinematics can be divided into two regions. For flexion angles less than 120°, the different muscle loads had a various affect on femoral translation and tibial rotation (data not shown here) [57]. Muscle load type also affected the forces in the cruciate ligaments below 120°. However, in high knee flexion, the application of muscle loads had little effect on knee kinematics or the forces in the ACL and PCL. In fact, muscle loads appear to play a minimal role in influencing maximal knee flexion. This suggests that the knee is highly constrained in high flexion.

Hefzy et al [39] recorded that at deep knee flexion (157°±4.9°) the lateral condyle translated more posteriorly than the medial condyle. The data presented here is in agreement with the published data. Researchers noticed [11, 12, 39, 40] that at high knee flexion, the tibial shaft contacted the most proximal aspect of the posterior condyles. In the intact knee, the medial meniscus follows the tibia closely, providing surface area for the contact with the medial femoral condyle. Interestingly, the lateral meniscus follows the tibia closely for flexion angles less than 120°, but beyond 120°, it extends beyond the tibial plateau to provide an "extended" surface for the femur to rollback over the tibia.

This additional "elongation" is one of the factors that provide the stability of the intact knee at high knee flexion.

The role of the soft tissue compression comes into play only in high knee flexion. As the knee flexes beyond 120°, the posterior soft tissue, including posterior capsule, menisci, hamstring muscles, skin, and fat, of the knee is compressed between the femoral shaft the posterior edge of the tibia. We believe that this posterior soft tissue compression plays an important role in promoting further posterior femoral translation in high knee flexion. This becomes possible due to the fact that the posterior soft tissue can no longer be compressed therefore, pushing the tibia anteriorly.

In conclusion, at lower flexion angles, the knee joint stability is maintained by the articular contact (including the menisci), muscle loads as well as the tension of the surrounding soft tissues. At high flexion, the muscles had little effect on knee kinematics. There is a strong resistance and constraint in high knee flexion. Posterior tibial translation may be limited by the impingement of the posterior soft tissues of the knee between the tibia and femur. Anterior tibial translation may be limited by a highly congruent articulation between the posterior horns of the menisci and the concave aspect of the proximal segment of the posterior condyles of the femur. The posterior edge of the tibia, together with the menisci, articulates with the concave surface of the condyle in high flexion. This provides a stable interaction of the tibiofemoral joint in high flexion and may prevent anterior subluxation of the knee.

4.6 References

1. Aglietti, P., R. Buzzi, and A. Gaudenzi, *Patellofemoral functional results and complications with the posterior stabilized total condylar knee prosthesis.* J Arthroplasty, 1988. 3(1): p. 17-25.
2. Dennis, D., et al., *Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation.* Clin Orthop, 1992. 281: p. 168-176.
3. Emmerson, K.P., C.G. Moran, and I.M. Pinder, *Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up.* J Bone Joint Surg Br, 1996. 78(3): p. 441-5.
4. Goldberg, V.M., et al., *Use of a total condylar knee prosthesis for treatment of osteoarthritis and rheumatoid arthritis. Long-term results.* J Bone Joint Surg Am, 1988. 70(6): p. 802-11.
5. Insall, J.N., et al., *The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements.* J Bone Joint Surg Am, 1983. 65(5): p. 619-28.
6. Insall, J.N., et al., *Total knee arthroplasty.* Clin Orthop, 1985(192): p. 13-22.
7. Lee, J.G., et al., *Review of the all-polyethylene tibial component in total knee arthroplasty. A minimum seven-year follow-up period.* Clin Orthop, 1990(260): p. 87-92.
8. Malkani, A.L., et al., *Total knee arthroplasty with the kinematic condylar prosthesis. A ten- year follow-up study.* J Bone Joint Surg Am, 1995. 77(3): p. 423-31.

9. Rand, J.A., *Comparison of metal-backed and all-polyethylene tibial components in cruciate condylar total knee arthroplasty.* J Arthroplasty, 1993. 8(3): p. 307-13.

10. Ranawat, C.S., C.P. Luessenhop, and J.A. Rodriguez, *The press-fit condylar modular total knee system. Four-to-six-year results with a posterior-cruciate-substituting design.* J Bone Joint Surg Am, 1997. 79(3): p. 342-8.

11. Li, G., et al., *Improving flexion in total knee arthroplasty*, in *The Adult Knee*, J. Callaghan, et al., Editors. 2002, Lippincott Williams & Wilkins: Philadelphia. p. 1233-1244.

12. Sultan, P., et al., *Optimizing Flexion After Total Knee Arthroplasty: Advances in Prosthetic Design.* Clin Orthop, 2003.

13. Mulholland, S.J. and U.P. Wyss, *Activities of daily living in non-Western cultures: range of motion requirements for hip and knee joint implants.* Int J Rehabil Res, 2001. 24(3): p. 191-8.

14. Nagura, T., et al., *Mechanical loads at the knee joint during deep flexion.* J Orthop Res, 2002. 20(4): p. 881-6.

15. Nakagawa, S., et al., *Tibiofemoral movement 3: full flexion in the living knee studied by MRI.* J Bone Joint Surg Br, 2000. 82(8): p. 1199-200.

16. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

17. Todo, S., et al., *Anteroposterior and rotational movement of femur during knee flexion.* Clin Orthop, 1999(362): p. 162-70.

18. Livesay, G.A., et al., *Evaluation of the effect of joint constraints on the in situ force distribution in the anterior cruciate ligament.* J Orthop Res, 1997. 15(2): p. 278-84.

19. Li, G., et al., *The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL.* J Biomech, 1999. 32(4): p. 395-400.

20. Li, G., et al., *Effect of combined axial compressive and anterior tibial loads on in situ forces in the anterior cruciate ligament: a porcine study.* J Orthop Res, 1998. 16(1): p. 122-7.

21. Hollister, A.M., et al., *The axes of rotation of the knee.* Clin Orthop, 1993(290): p. 259-68.

22. Hirokawa, S., et al., *Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction.* Am J Sports Med, 1992. 20(3): p. 299-306.

23. Durselen, L., L. Claes, and H. Kiefer, *The influence of muscle forces and external loads on cruciate ligament strain.* Am J Sports Med, 1995. 23(1): p. 129-36.

24. Draganich, L.F., T.P. Andriacchi, and G.B. Andersson, *Interaction between intrinsic knee mechanics and the knee extensor mechanism.* J Orthop Res, 1987. 5(4): p. 539-47.

25. Komistek, R.D., D.A. Dennis, and M. Mahfouz, *In vivo fluoroscopic analysis of the normal human knee.* Clin Orthop, 2003(410): p. 69-81.

26. Beynnon, B.D., et al., *Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo.* Am J Sports Med, 1995. 23(1): p. 24-34.

27. Arms, S.W., et al., *The biomechanics of anterior cruciate ligament rehabilitation and reconstruction.* Am J Sports Med, 1984. 12(1): p. 8-18.

28. Andriacchi, T. and J. Galante, *Retention of the posterior cruciate in total knee arhtroplasty.* J Arthroplasty, 1988. Supplement: p. S13-S19.

29. Iwaki, H., V. Pinskerova, and M.A. Freeman, *Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee.* J Bone Joint Surg Br, 2000. 82(8): p. 1189-95.

30. Hill, P.F., et al., *Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI.* J Bone Joint Surg Br, 2000. 82(8): p. 1196-8.

31. Li, G., et al., *Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads--an in vitro experimental study.* J Orthop Res, 2002. 20(4): p. 887-92.

32. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.

33. Most, E., et al., *Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty.* Clin Orthop, 2003(410): p. 101-13.

34. Most, E., et al., *The Kinematics of Fixed and Mobile Bearing Posterior Stabilized Total Knee Arthroplasty.* Clin Ortho, 2003. 416: p. 1-11.

35. Freeman, M.A. and V. Pinskerova, *The movement of the knee studied by magnetic resonance imaging.* Clin Orthop, 2003(410): p. 35-43.

36. Dennis, D.A., et al., *In vivo knee kinematics derived using an inverse perspective technique.* Clin Orthop, 1996(331): p. 107-17.

37. Asano, T., et al., *In vivo three-dimensional knee kinematics using a biplanar image- matching technique.* Clin Orthop, 2001(388): p. 157-66.

38. Piazza, S.J. and P.R. Cavanagh, *Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment.* J Biomech, 2000. 33(8): p. 1029-34.

39. Hefzy, M.S., B.P. Kelly, and T.D. Cooke, *Kinematics of the knee joint in deep flexion: a radiographic assessment.* Med Eng Phys, 1998. 20(4): p. 302-7.

40. Spanu, C. and M. Hefzy, *Biomechanics of the knee joint un deep felxion: A prelude to a total knee replacement that allow for maximum flexion.* Technology and Health Care, 2003. 11: p. 161-181.

41. Butler, D.L., F.R. Noyes, and E.S. Grood, *Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study.* J Bone Joint Surg Am, 1980. 62(2): p. 259-70.

42. Fanelli, G.C., B.F. Giannotti, and C.J. Edson, *The posterior cruciate ligament arthroscopic evaluation and treatment.* Arthroscopy, 1994. 10(6): p. 673-88.

43. Fox, R.J., et al., *Determination of the in situ forces in the human posterior cruciate ligament using robotic technology. A cadaveric study.* Am J Sports Med, 1998. 26(3): p. 395-401.

44. Gollehon, D.L., P.A. Torzilli, and R.F. Warren, *The role of the posterolateral and cruciate ligaments in the stability of the human knee. A biomechanical study.* J Bone Joint Surg Am, 1987. 69(2): p. 233-42.

45. Hoher, J., et al., *In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study.* J Orthop Res, 1999. 17(5): p. 763-8.

46. Markolf, K.L., et al., *A biomechanical study of replacement of the posterior cruciate ligament with a graft. Part 1: Isometry, pre-tension of the graft, and anterior-posterior laxity.* J Bone Joint Surg Am, 1997. 79(3): p. 375-80.

47. Markolf, K.L., et al., *A biomechanical study of replacement of the posterior cruciate ligament with a graft. Part II: Forces in the graft compared with forces in the intact ligament.* J Bone Joint Surg Am, 1997. 79(3): p. 381-6.

48. Vogrin, T.M., et al., *Effects of sectioning the posterolateral structures on knee kinematics and in situ forces in the posterior cruciate ligament.* Knee Surg Sports Traumatol Arthrosc, 2000. 8(2): p. 93-8.

49. Rudy, T.W., et al., *A combined robotic/universal force sensor approach to determine in situ forces of knee ligaments.* J Biomech, 1996. 29(10): p. 1357-60.

50. Hefzy, M.S., et al., *Knee kinematics in-vivo of kneeling in deep flexion examined by bi-planar radiographs.* Biomed Sci Instrum, 1997. 33: p. 453-8.

51. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

52. Andriacchi, T.P., *Functional analysis of pre and post-knee surgery: total knee arthroplasty and ACL reconstruction.* J Biomech Eng, 1993. 115(4B): p. 575-81.

53. Kanamori, A., et al., *The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology.* Arthroscopy, 2000. 16(6): p. 633-9.

54. Markolf, K.L., et al., *Biomechanical consequences of replacement of the anterior cruciate ligament with a patellar ligament allograft. Part II: forces in the graft*

compared with forces in the intact ligament. J Bone Joint Surg Am, 1996. 78(11): p. 1728-34.

55. Markolf, K.L., et al., *Biomechanical consequences of replacement of the anterior cruciate ligament with a patellar ligament allograft. Part I: insertion of the graft and anterior-posterior testing.* J Bone Joint Surg Am, 1996. 78(11): p. 1720-7.

56. Zayontz, S., et al. *ACL and PCL do not carry significant loads in high flexion of the knee under muscle loads*. in *Trans Orthop Res So*. 2002. Dallas.

57. Li, G., et al., *Kinematics of the knee at high flexion angles: an in-vitro investigation.* J Orthop Res, 2004.

Chapter 5

THE EFFECT OF THE POSTERIOR CRUCIATE LIGAMENT ON TOTAL KNEE ARTHROPLASTY

5.1 Introduction

The role of the posterior cruciate ligament (PCL) in total knee arthroplasty (TKA) has been studied and debated for many years [1-8]. It has been argued that retaining the PCL in TKA can enhance joint stability, improve passive range of motion by allowing femoral rollback, increase efficacy of the knee musculature, and reduce stress at the implant-cement and cement-bone interfaces [1, 9, 10]. Furthermore, the PCL can contribute to the posterior stability in a flexed knee provided that its tension can be accurately restored after TKA [11]. However, normal tension of the PCL is often difficult to achieve through soft tissue balancing after TKA [3]. Follow-up studies of patients after TKA found that partial release of the PCL is beneficial for patients with tight PCL at the time of knee arthroplasty [8, 12, 13].

Many studies investigated knee kinematics after TKA [2, 14-19]. Matsuda et al [15] measured the anterior-posterior laxity of a PCL-retaining TKA (Miller-Galante 1, Zimmer, Warsaw, IN) using a KT-2000 arthrometer and found inconsistent anterior-posterior stability in flexion among 19 knees. Stiehl et al [17, 18] studied various PCL-retaining knee designs using a fluoroscopic technique and found that physiological rollback of the femur was not demonstrated in patients after PCL-retaining TKA. Similar results were also observed by Kim et al [19] in PCL-retaining TKA (Genesis, Smith and Nephew, Richards Inc, Memphis, TN). Dennis et al [2] found abnormal femoral translation during deep knee-bends in patients after TKA (Press-Fit Condylar Designs, Johnson & Johnson, Raynham, MA). Furthermore, Dennis et al [20] demonstrated that PCL-retaining TKA has a similar range of passive motion to PCL-substituting TKA but a decreased range of motion during squatting. Using similar fluoroscopic technique, however, Banks et al [16] found that the range of axial tibial rotation and condylar translation for PCL-retaining TKA (AMK, DePuy, Warsaw, IN) was similar to the range reported for normal and anterior cruciate ligament (ACL)-deficient knees during a step-up maneuver. Tarnowski et al [14] found that the anterior-posterior tibial displacement of the PCL-retaining TKA (NexGen, Zimmer, Warsaw, IN) was not significantly lower than that of the normal knee during gait.

The capabilities of the PCL-retaining TKA to restore normal knee kinematics and function remain controversial. Most research to date has focused on the measurement of TKA kinematics during a specific activity. However, quantitative evaluation of TKA function has been difficult since intact knee behavior is usually unavailable in both *in vivo* and *in vitro* studies. In addition, to our knowledge, no data has been reported on the *in situ* forces of the PCL in TKA during functional activities, and no study has correlated PCL forces with TKA kinematics and contact mechanics.

This chapter contains data from two experiments whereby the robotic testing system was utilized to study the effect of the PCL on the biomechanics of PCL-retaining TKAs:

Study 1: In this study, the kinematics (femoral translation and tibial rotation) of a single-design PCL-retaining TKA (NexGen CR, Zimmer, Warsaw, IN) was determined and compared with that of the intact knee under various muscle loads. The effect of PCL resection on PCL-retaining TKA kinematics was explored. In addition, the *in situ* forces of the PCL in the TKA were quantitatively measured. This study was limited to flexion range between 0° and 120° as per manufacture specifications.

Study 2: A new PCL-retaining TKA design was introduced (NexGen CR-Flex, Zimmer Inc., Warsaw, IN) to potentially enhance further range of motion beyond 120°. The CR-Flex design included 2mm extended femoral condyles compared to the conventional PCL-retaining (NexGen CR, Zimmer Inc., Warsaw, IN) design to allow larger tibiofemoral contact area at high flexion angles. No information is available regarding the kinematics advantages of this high flexion PCL-retaining TKA design. In this study, we compared the femoral translation and tibial rotation of intact knee, conventional PCL-retaining (NexGen CR) TKA, and high flexion PCL-retaining (CR-Flex) TKA designs from full extension to full flexion (150°). The PCL force in the high flexion PCL-retaining TKA as a function of knee flexion was investigated and compared to the intact knee PCL force (chapter 4) and PCL force in the conventional design (study1).

Additionally, the contact patterns (contact point location, and contact area) of these two CR TKAs components were compared throughout the entire range of flexion.

The chapter provides complete kinematics, kinetics, and contact mechanics information that is useful in understanding the limiting factors posterior-retaining TKA that prevent patients from flexing beyond 120°.

5.2 Experimental Set-Up

5.2.1 Study 1

5.2.1.1 Specimen Preparation

Nine fresh-frozen cadaveric human knee specimens were tested using a robotic testing system. These specimens were 66 ± 6 (mean ± SD) years old and displayed no radiological evidence of osteoarthritis. Furthermore, each specimen had a full range of motion and intact ligaments. Each specimen was thawed at room temperature overnight before testing. The femur and tibia were each cut to approximately 25 cm in length from the joint line, and the bones were secured in thick-walled aluminum cylinders. Soft tissues around the knee joint (including ligaments, muscles and skin) were left intact.

Prior to mounting the specimen onto the robot, nylon cables were sutured onto the tendons of the quadriceps and hamstring muscles using Ethibond #5 polyester suture. The tendon of the quadriceps was identified above the knee joint; the medial hamstrings tendons, comprised of the semitendinosus and semimembranosus, were sutured *en masse*; and, finally, the tendon of biceps femoris was identified and sutured to the cable. During installation, the cables were carefully aligned with the long axis of the femur. The femur was rigidly fixed to a base, and the tibia was fixed to the robotic arm through a six-degrees-of-freedom force-moment sensor. The robotic manipulator, together with the force-moment sensor, could learn the complex motion of the knee specimen in response to external loads and could then reproduce these motions in sequential tests.

5.2.1.2 Coordinate System Determination

After the specimen was installed onto the robotic testing system in full extension, a MicroScribe® digitization system (Microscribe 3DX, Immersion Technology, San Jose, CA) was used to digitize landmarks on the specimen to determine a knee coordinate system as previously described (Chapter 2).

5.2.1.3 Kinematics Analysis

At each flexion angle of the knee, a *passive position* was determined by the robotic testing system, where the knee carried minimal load (below 5 N in forces and 0.5 Nm in moments). The passive path was defined by a series of passive positions of the knee from full extension to 120° (in 1° increments) [21-23]. This path of least resistance, analogous to passive movement in the living knee, was recorded by the robot and used as a reference (or starting) position when applying external loads.

After determining the passive path, the knee was tested at selected flexion angles: 0°, 30°, 60°, 90° and 120°. At each of these angles, a series of muscle loads were applied to the knee using hanging weights onto the cables to simulate isometric flexion and extension exercises of the knee [24]. Peak quadriceps load can reach three to six times body weight during *in vivo* activities [25]. However, due to testing limitations, the following forces were applied in this experiment: (1) isolated quadriceps forces (400 N, approximately one half bodyweight), (2) combined quadriceps/hamstrings load (400

N/200 N with 100N for each of the medial and lateral hamstrings), and (3) isolated hamstrings forces (200 N).

At each of the above flexion angles, the knee was placed at its passive position, and muscle loads were applied. The robot moved the tibia in the remaining five degrees-of-freedom until reaching an equilibrium position where the internal resistance of the knee balanced the applied muscle load. This position represented the kinematics response of the knee to the muscle load.

5.2.1.4 Knee Reconstruction

The intact knee was then replaced with a PCL-retaining TKA (NexGen CR, Zimmer, Warsaw, IN). This design has an asymmetric femoral component. The radii of the lateral condyle are larger than those of the medial condyle in the sagittal plane in order to facilitate axial rotation of the knee during flexion. The tibiofemoral articulating surface radii are matched to provide conforming surfaces in the coronal plane to increase the contact area. The patellar groove was deepened and made asymmetric to improve patellofemoral tracking. The tibial component of this prosthesis has a curved articular surface and a posterior lip. Intramedullary alignment was used on the femoral side. The femur was cut in 6° of valgus and 3° of external rotation. The epicondylar axis was used to assess the rotational alignment, with the posterior femoral condyles and Whiteside's line as additional references. Tibial alignment was performed using an extramedullary guide. The reference points used were the center of the tibial plateau, the junction of the medial and middle thirds of the tibial tuberosity and the visible part of the tibial crest. The tibial cut was performed with a 7° posterior slope. A bone block was created posteriorly to protect the PCL insertion. Prior to placement of the definitive components, a trial reduction was performed with careful attention to assess the flexion and extension gaps, stability, range of motion and patellar tracking. PCL tension was assessed by flexing and extending the knee and examining for tibial tray anterior lift-off. The PCL was also manually palpated to assess for tension, and the flexion gap was examined. The PCL was partially recessed from the tibial insertion in only one knee because of a tight flexion space [12]. The femoral prosthesis was inserted without cement in a secure press-fit manner; the tibial component was cemented. The patella was not resurfaced in this study. The extensor mechanism and skin were closed with sutures in a standard fashion.

The reconstructed knee was then tested using the robotic testing system. A new passive path of the TKA was determined, analogous to the passive motion of the patient after TKA. Flexion positions (0°, 30°, 60°, 90°, and 120°) on this path were used as the initial position for applying external loads to the knee after TKA. The same muscle loads that were applied to the intact knee were also applied to the TKA at the same flexion angles. The robot measured the corresponding kinematics of the PCL-retaining TKA, and the load cell recorded the forces transferred through the joint.

5.2.1.5 Measurements of PCL Force in Conventional PCL-Retaining TKA

To measure the *in situ* forces of the PCL in the TKA, the PCL was then transected via the previous arthrotomy. The kinematics of the TKA measured before cutting the PCL and under the muscle loads were reproduced by the robot on the specimen, and the load cell recorded the forces transferred through the joint. The *in situ* forces of the PCL in the TKA were then obtained by calculating the differences of the forces measured before and after the PCL resection using the principle of superposition [24, 26]. The PCL forces measured in the TKA were compared to those of intact knees under the same loading conditions, but in different specimens.

5.2.1.6 PCL-Deficient TKA

Complete transection of the PCL resulted in a PCL-deficient TKA. The same testing procedure used for the PCL-retaining TKA was then used to test the PCL-deficient TKA. The experimental data included a passive path and knee kinematics under the muscle loads at selected flexion angles. This allowed direct assessment of the contribution of the PCL to TKA kinematics by comparing the results of the TKA with and without the PCL in the same knee.

5.2.1.7 Statistical Analysis

A two-way repeated measure Analysis of Variance (ANOVA) was used to detect whether the knee state (i.e., intact knee, PCL-retaining TKA and PCL-deficient TKA) had a significant affect on knee kinematics (femoral translation and tibial rotation). A one-way non-repeated measure Analysis of Variance (ANOVA) was used to detect statistical differences between *in situ* intact PCL forces and PCL forces in conventional PCL-retaining TKA under the simulated muscle loads. Statistical significance was set as $p < 0.05$.

5.2.2 Study 2

5.2.2.1 Specimen Preparation

Ten fresh-frozen post-mortem human cadaveric knees (2 male knees and 8 female knees with average age 74±15 years; age range 48-96) were used in this study. Each specimen was approximately 50cm long (25cm from the joint line of femur and tibia).

Each specimen was thawed over night and radiographed to exclude specimens with previous injuries. The muscles, ligaments, capsule, skin, and menisci remained intact around the knee joint. All knees were tested to assure full range of motion (from full extension to 150° of flexion). Specimens that could not achieve full range of flexion (0° to 150°) by manual flexion were excluded from the study. The fibula was secured to the tibial shaft in its anatomical position using a cortical screw. The specimen preparation and installation on the robotic testing system was similar to study 1.

5.2.2.2 Data Analysis

The coordinate system is described in detail in chapter 2 (Section 2.2). The anteroposterior translation of the medial and lateral centers on the transepicondylar axis represented femoral translations. All measurements were reported with respect to the knee at full extension. Internal and external tibial rotations were calculated using the convention as defined by Grood and Santay [27].

5.2.2.3 Kinematics Analysis

Each specimen was manually preconditioned ten times prior to its installation on the robotic testing system [21-23, 28]. The intact knee was tested first. Its passive path was determined and served as a baseline for all other tests. The kinematics (femoral translation and tibial rotation) under combined muscle loads (400N quadriceps and 200N hamstrings) were then determined at selected flexion angles (0°, 30°, 60°, 90°, 120°, 135°, and 150°). An orthopaedic surgeon then performed an arthroplasty surgery on the intact knee to fit a conventional PCL-retaining TKA (NexGen CR, Zimmer Inc., Warsaw, IN). A new passive path and a new knee kinematics for the PCL-retaining TKA under the same loadings were determined. Next, the femoral and polyethylene components were changed to fit a high flexion PCL-retaining TKA (NexGen CR-Flex, Zimmer Inc., Warsaw, IN). The same testing protocol was repeated for the modified knee. A new passive path and a new knee kinematics for the high flexion PCL-retaining TKA under the same muscle loading were determined.

5.2.2.4 PCL Force Measurements in High Flexion PCL-Retaining TKA

After measuring the kinematics of the high flexion PCL-retaining TKA, the PCL was excised completely and the pre-recorded six degrees-of freedom kinematics of the high flexion PCL-retaining TKA was replayed. The load cell measured the force within the knee joint while replaying the kinematics. The PCL force in the high flexion PCL-retaining TKA was obtained using the principle of superposition [22, 29, 30].

5.2.2.5 Contact Mechanics Assessment

After determining TKA kinematics for the conventional and high flexion PCL-retaining TKA designs, all soft tissue around the knee joint was removed. Pressure sensor (K-scan 4000, Tekscan® Inc., South Boston, MA) was inserted into the articulation. The sensor edge was aligned with the posterior edge of the polyethylene to assure consistency among all specimens (Figure 20). A double sided tape was used to hold the sensor in its place to prevent slippage throughout the entire experiment. Prerecorded TKA kinematics were then replayed in series, using the appropriate components. The corresponding contact areas and peak contact pressure point were recorded at selected flexion angles (0°, 30°, 60°, 90°, 120°, 135°, and 150°) for NexGen CR and CR-Flex designs.

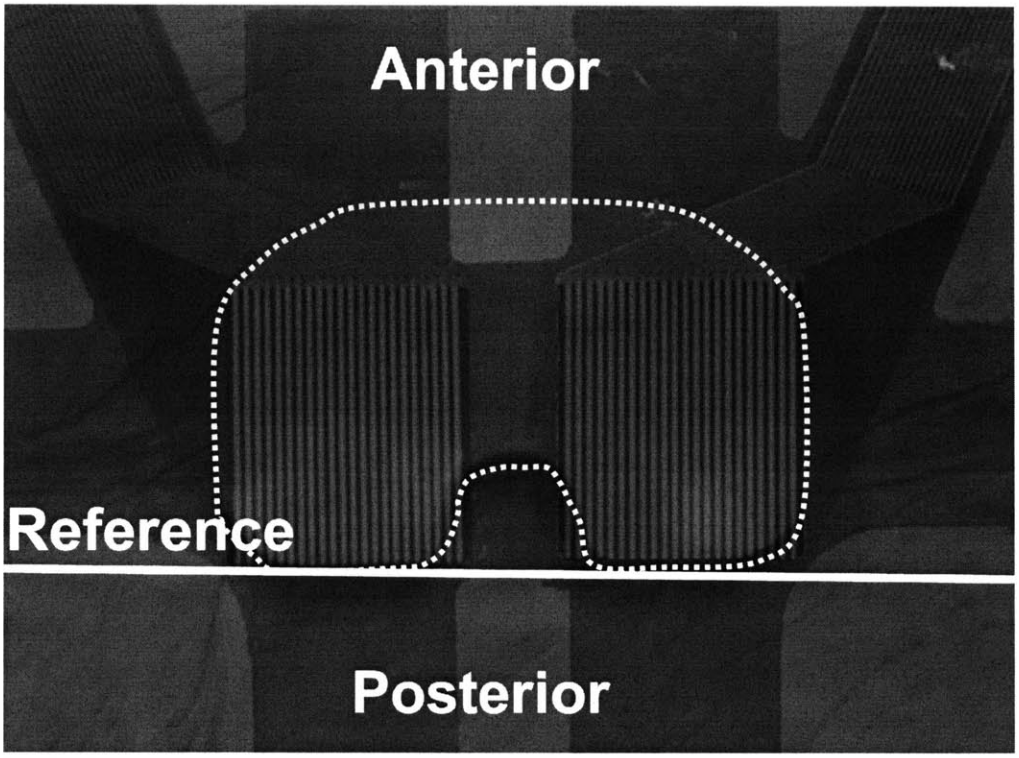


Figure 20: The position of the TekScan® film over the polyethylene liner.

5.2.2.6 Statistical Analysis

One- and two-way repeated measures Analysis of Variance (ANOVA) were performed to detect whether the knee state had a significant effect on femoral translation and tibial rotation. Newman-Keuls tests were done to detect statistical significance between knee states. Student t-test was used to detect statistical differences between conventional and high flexion CR design in terms of contact area and contact location. The significance level was set to a p-value less than 0.05.

5.3 Results

5.3.1 The Effect of the PCL on Knee Kinematics (femoral translation and tibial rotation) – Study 1

5.3.1.1 Femoral Translation and Tibial Rotation on the Passive Path – Study 1

The intact knee at full extension under no load condition served as a reference for all tests. With increasing knee flexion, the femur of the intact knee gradually translated posteriorly reaching a value of 12.2 ± 4.7 mm (mean ± std) at 90° and a peak value of 14.9 ± 6.9 mm at 120° (Figure 21). The femoral condyle of the PCL-retaining TKA was posteriorly positioned at full extension when compared to the intact knee (Figure 21). It then translated slightly in an anterior direction (not statistically significant) at 30° (-0.2 ± 4.0). Thereafter, it translated posteriorly with flexion reaching a value of 6.6 ± 5.2 mm at 90°. At 120°, it reached maximal posterior translation (9.4±5.8 mm) which was approximately 63% of that seen in the intact knee. The absence of the PCL significantly changes knee kinematics for flexion angles greater that 30° (Figure 21). At full extension, the femoral condyle of the conventional PCL-deficient TKA was situated posteriorly at 0.4 ± 2.3 mm; afterward, it translated anteriorly reaching a peak anterior translation of -1.6 ± 5.0 mm at 30°. With further increase in knee flexion, the femoral condyle translated more posteriorly reaching a peak posterior value of 3.9 ± 5.0 mm at 120°. The PCL-deficient TKA was statistically different from the intact knee for all flexion angles greater than 0°. At 90° and 120°, the PCL-deficient TKA was also statistically different from the PCL-retaining TKA.

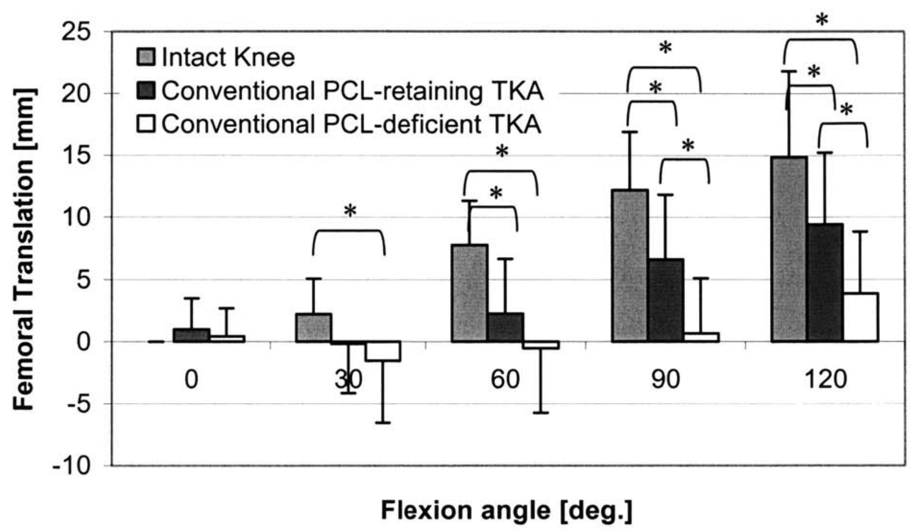


Figure 21: Posterior (+) femoral translation (knee center) on the passive path for intact, conventional PCL-retaining, and conventional PCL-deficient TKAs as a function of flexion angles (* denotes $p < 0.05$).

For all knees, internal tibial rotation increased with increasing flexion angle (Figure 22). In the intact knee at 30° of flexion, the tibia internally rotated 4.8° ± 3.2°, reaching a maximum of 8.3° ± 11.7° at 120° flexion. After reconstruction using the PCL-retaining TKA, the tibia rotated internally 3.7° ± 6.4° at full extension and the amount of rotation remained relatively unchanged for flexion angles less than 60°. Thereafter, internal tibial rotation increased gradually with increasing flexion angle reaching a maximum value of 11.2° ± 15.2° at 120°. A similar trend was noted for the knee after resection of the PCL (PCL-deficient TKA). Internal tibial rotation was relatively constant for flexion angles less than 60° (<5°); afterward, it increased reaching a maximum value of 12.6° ± 12.1° at 120°. There was no significant difference in terms of the amount of internal tibial rotation between the intact knee and both TKAs throughout the range of passive knee motion (p>0.05).

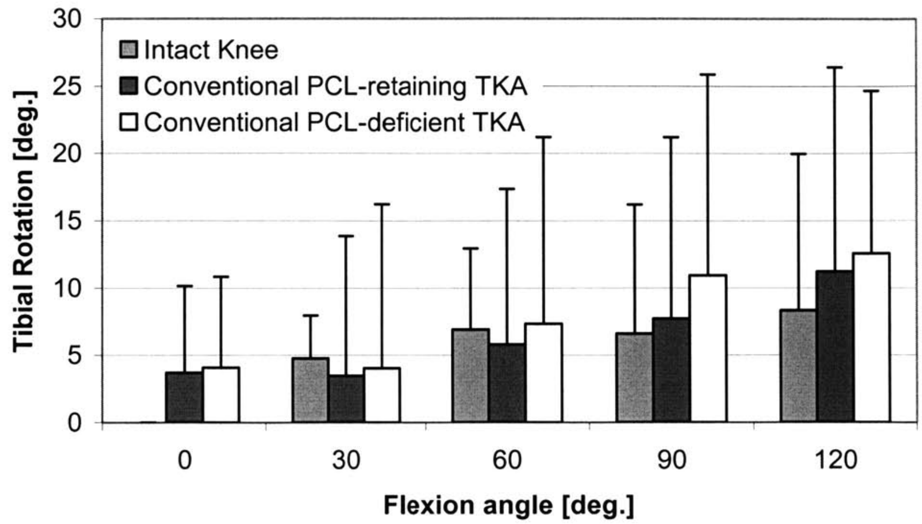


Figure 22: Femoral translation (knee center) on the passive path for intact, conventional PCL-retaining, and conventional PCL-deficient TKAs as a function of flexion angles (* denotes p < 0.05).

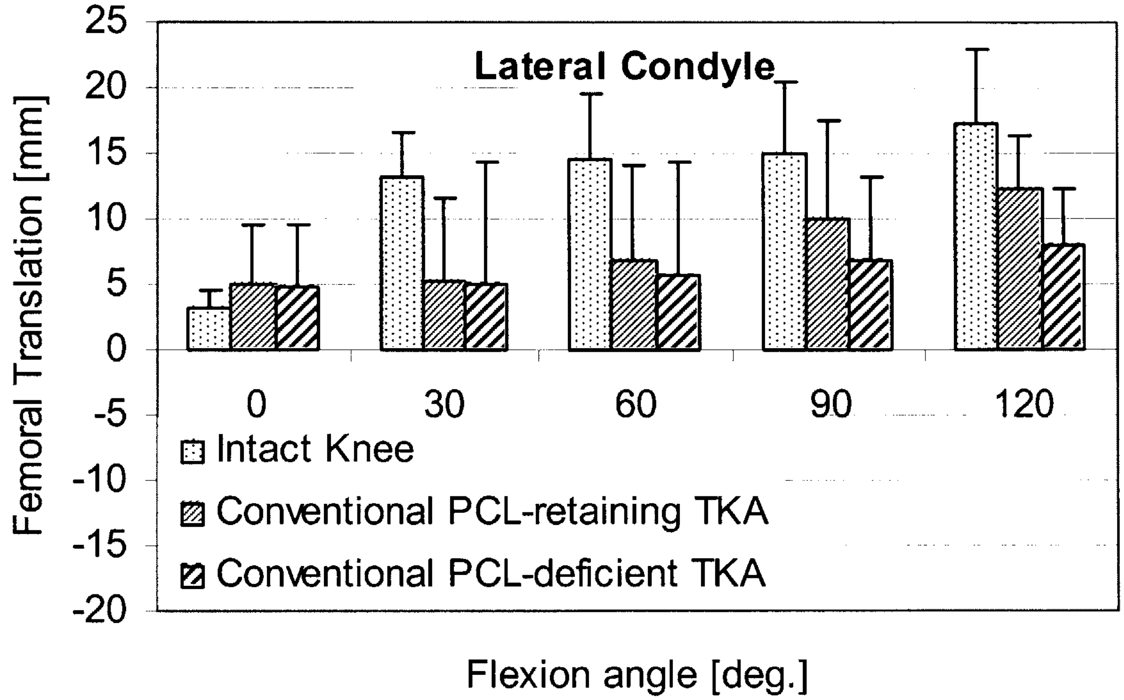
5.3.1.2 Femoral Translation under Various Muscle loads – Study 1

Posterior translations of lateral and medial femoral condyles of the reconstructed knee were different from those of the intact knee beyond full extension under the various muscle loads (Figures 23-25). The PCL-retaining TKA showed lower posterior femoral translation than the intact knee beyond 30° of flexion both medially and laterally. PCL transection in the TKA further reduced the posterior femoral translation. For example, under combined quadriceps and hamstring loading, the lateral condyle of the intact knee translated posteriorly 12.2 ± 3.2 mm at 30° and 15.2 ± 6.4 mm at 120° (Figure 24A). After PCL-retaining TKA, the lateral condyle translated 3.5 ± 7.3 mm at 30° and 9.8 ± 5.4 mm at 120°, which was significantly lower (p<0.05) than the intact knee (71% and

36%, respectively). After the PCL was transected, the posterior femoral translation was not substantially affected from full extension to 60° of flexion but was significantly reduced ($p<0.05$) to 2.7 ± 7.7 mm and 5.4 ± 5.2 mm at 90° and 120° of flexion, respectively.

On the medial side, the femoral condyle of the intact knee moved posteriorly 1.9 ± 2.8 mm at 30° and 10.4 ± 11.7 mm at 120° (Figure 24B). After PCL-retaining TKA, the medial femoral condyle translated anteriorly -0.9 ± 6.4 mm at 30° and posteriorly 1.6 ± 12.5 mm at 120°, which was significantly lower ($p<0.05$) than in the intact knee beyond 30° of flexion. After the PCL was transected in the PCL-retaining TKA, no significant alteration in posterior femoral translation was observed between full extension and 60° of flexion. However, at 90° and 120°, the medial condyle translated anteriorly -4.6 ± 12.3 and -2.4 ± 10.4 mm, respectively, representing significant reductions posterior femoral translation compared with those before PCL transection.

A.



B.

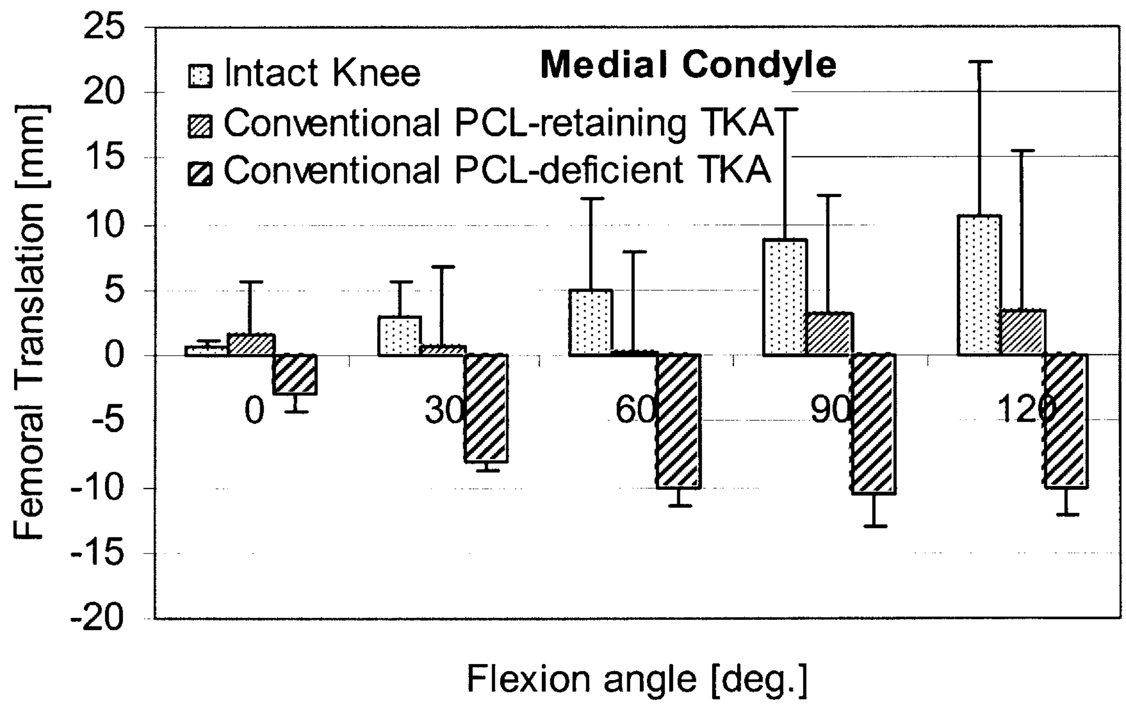


Figure 23: Posterior (+) femoral translation (femoral rollback) of the knee under isolated quadriceps load; A) lateral femoral condyle; and B) medial femoral condyle.

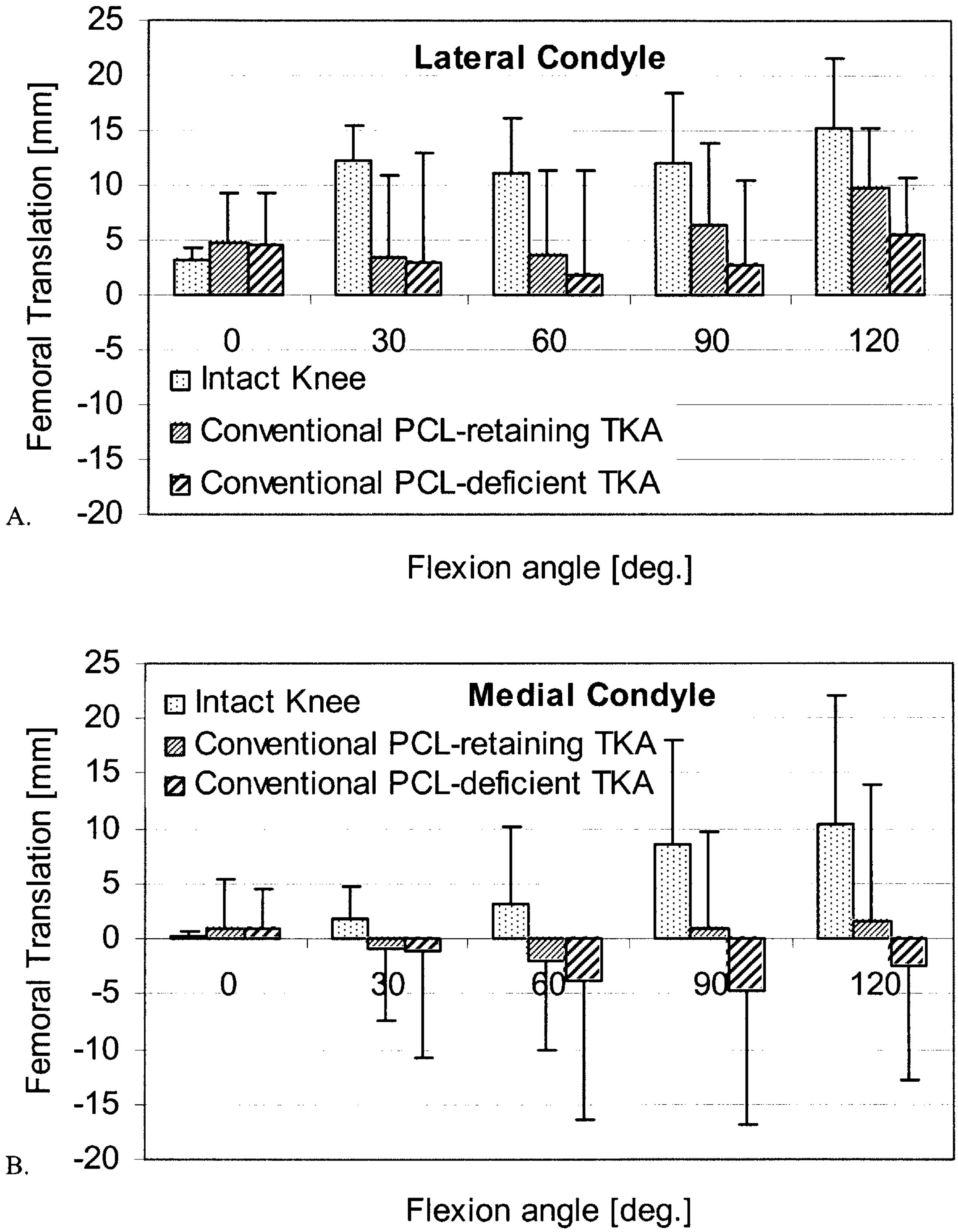
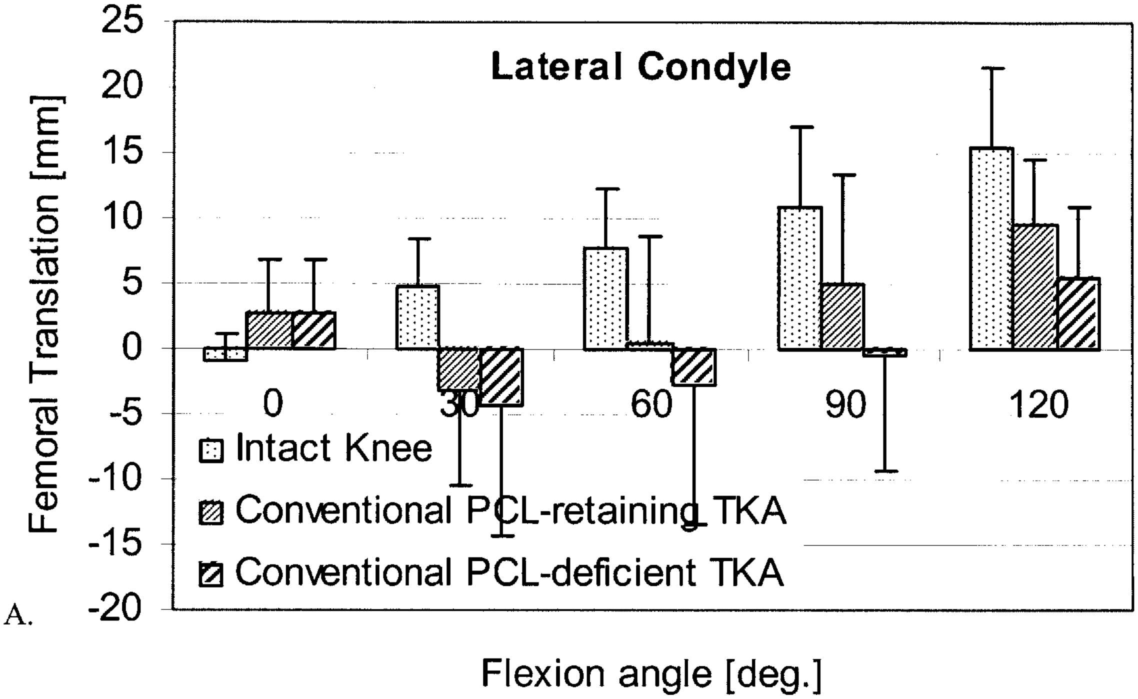
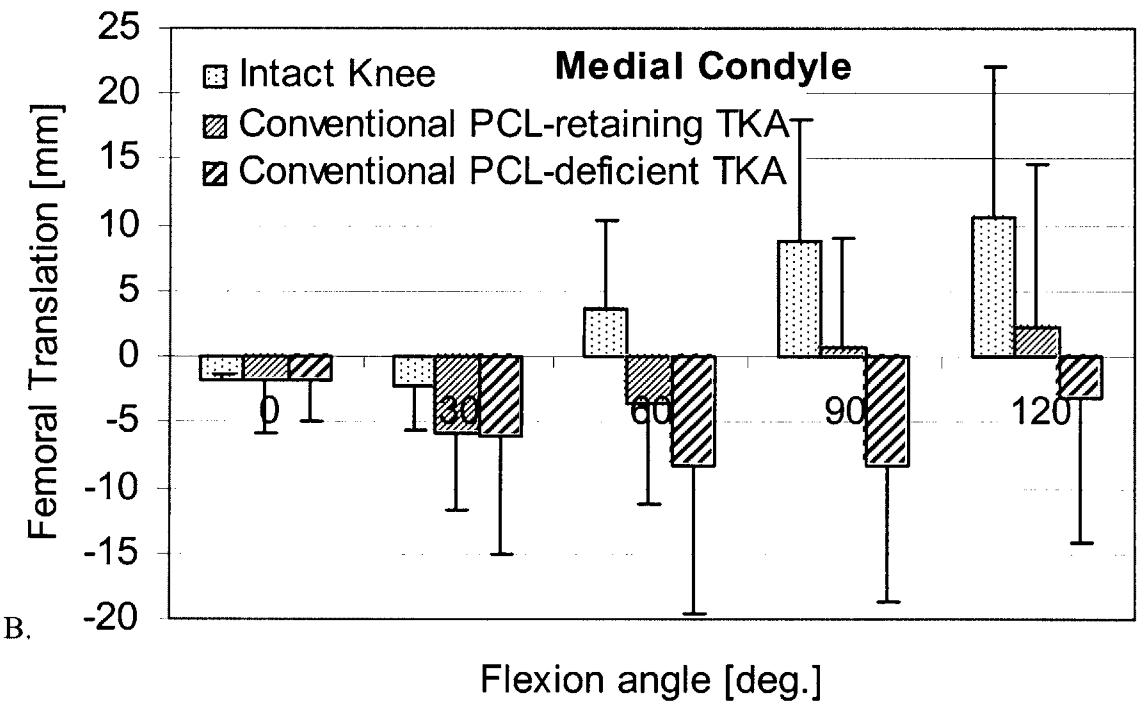


Figure 24: Posterior (+) femoral translation (femoral rollback) of the knee under the combined quadriceps and hamstring load; A) lateral femoral condyle, and B) medial femoral condyle.



A.

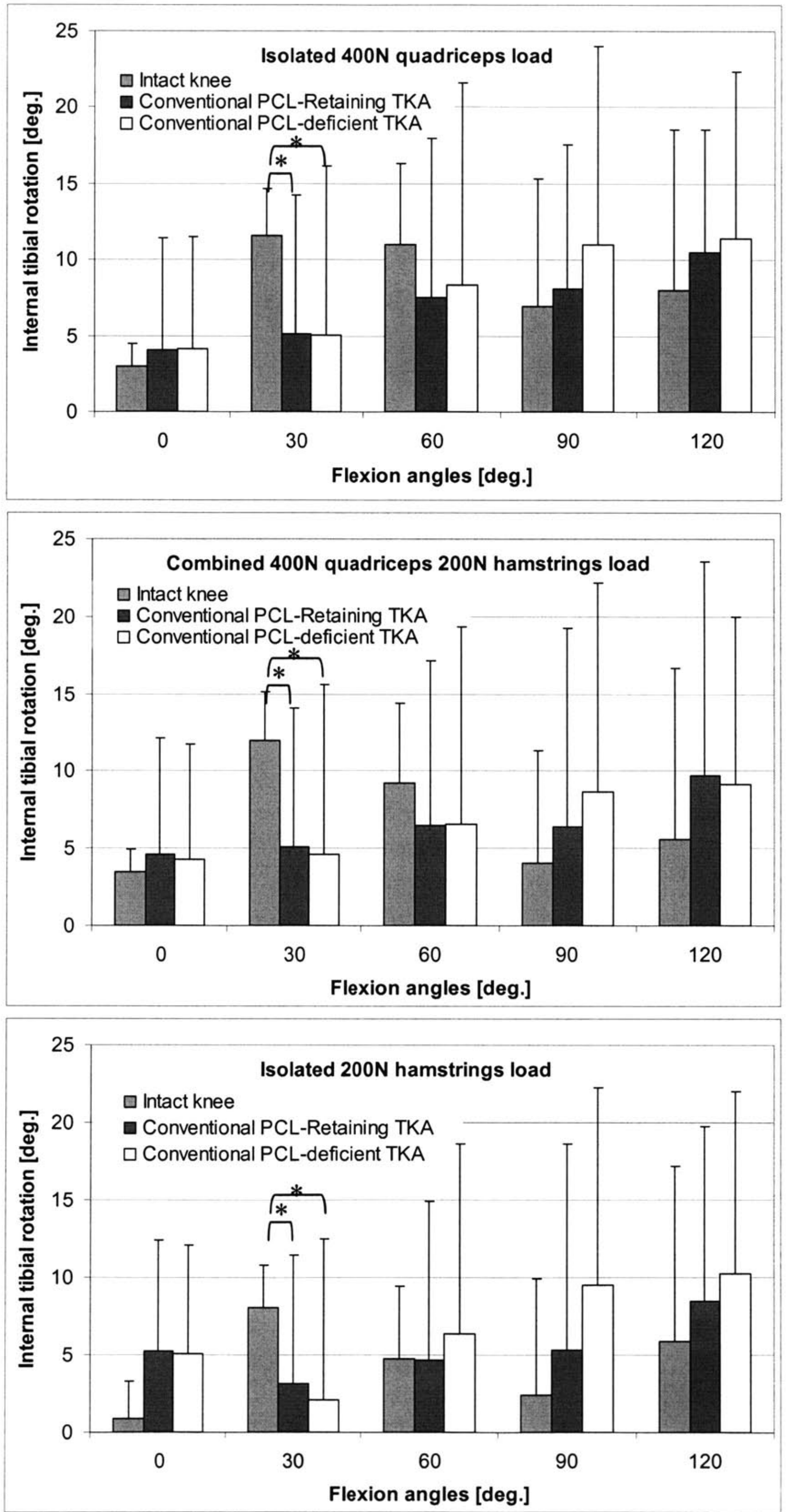


B.

Figure 25: Posterior (+) femoral translation (femoral rollback) of the knee under the isolated hamstring load; A) lateral femoral condyle, and B) medial femoral condyle.

5.3.1.3 Tibial Rotation under Various Muscle loads – Study 1

The intact knee, PCL-retaining, and PCL-deficient TKAs rotated internally throughout the range of flexion under simulated muscle loads (Figure 26A-C), but the rotation patterns of the intact knee and the TKAs were different. The internal tibial rotation of the intact knee reached peak values at 30° of flexion under all the muscle loads, while both the PCL-retaining and PCL-deficient TKAs showed peak internal tibial rotations at 120° of flexion. For example, under the combined quadriceps (400N) and hamstrings (200N) load, the tibia of the intact knee rotated 3.4 ± 1.5° internally at full extension (Figure 26B). The tibial rotation reached a peak value of 12.0 ± 3.1° at 30° of flexion. Thereafter, the internal tibial rotation decreased to 4.0 ± 7.3° at 90° and 5.6 ± 11.0° at 120° of flexion. However, the internal tibial rotation of the TKAs increased from full extension to 120° of flexion. The tibia of the PCL-retaining TKA rotated internally 4.6 ± 7.6° at full extension, 5.1 ± 8.9° at 30°, and increased to 9.7 ± 13.8° at 120° of flexion. The rotation of the PCL-retaining TKA was significantly lower ($p<0.05$) than that of the intact knee at 30° of flexion. The mean values of internal rotation of the TKA at 120° of flexion were higher than those of the intact knee under the muscle loads, even though the differences were not statistically significant. The PCL-deficient TKA also had a significantly lower ($p<0.05$) internal tibial rotation than that of the intact knee at 30° of flexion but had higher internal tibial rotation than the intact knee at high flexion angles of 90° and 120° of flexion (Figure 26B). There were no statistically significant differences between the PCL retaining and deficient TKAs in tibial rotation.



A.

B.

C.

Figure 26: Internal tibial rotation of the knee under various muscle loads: A) isolated quadriceps load, B) combined quadriceps and hamstrings load, and C) isolated hamstrings load (* $p<0.05$).

5.3.2 The PCL force in a conventional PCL-Retaining Total Knee Arthroplasty – Study 1

Under isolated quadriceps load of 400N, the average *in situ* forces of the PCL in the conventional PCL-retaining TKA were below 20 N at 0°, 30°, and 60° of flexion (Figure 27). At 90° the recorded PCL force was 18.3 ± 18.0 N. With further knee flexion, the PCL force increased reaching a peak value 28.8 ± 29.0 N at 120° of flexion. With a combined quadriceps and hamstrings load, the PCL forces were minimal (<20 N) for flexion angles less than 90°. The forces increased with increasing flexion reaching a value of 24.1 ± 19.5 N at 90° of flexion. The peak value of the PCL force under the combined quadriceps and hamstrings load was 33.5 ± 24.3 N at 120° of flexion (Figure 27).

Under isolated hamstrings load of 200 N, the PCL forces were less than 20 N at 0° and 30° of flexion, which were similar to those observed under the quadriceps loads (Figure 27). Beyond 30° of flexion, the hamstrings load caused higher PCL forces than all other muscle loads at 60°, 90° and 120° of flexion. Peak PCL force under isolated hamstrings load was 59.4 ± 28.7 N at 90° of flexion. Afterward, PCL force decreased to 43.4 ± 23.1 N at 120° of flexion.

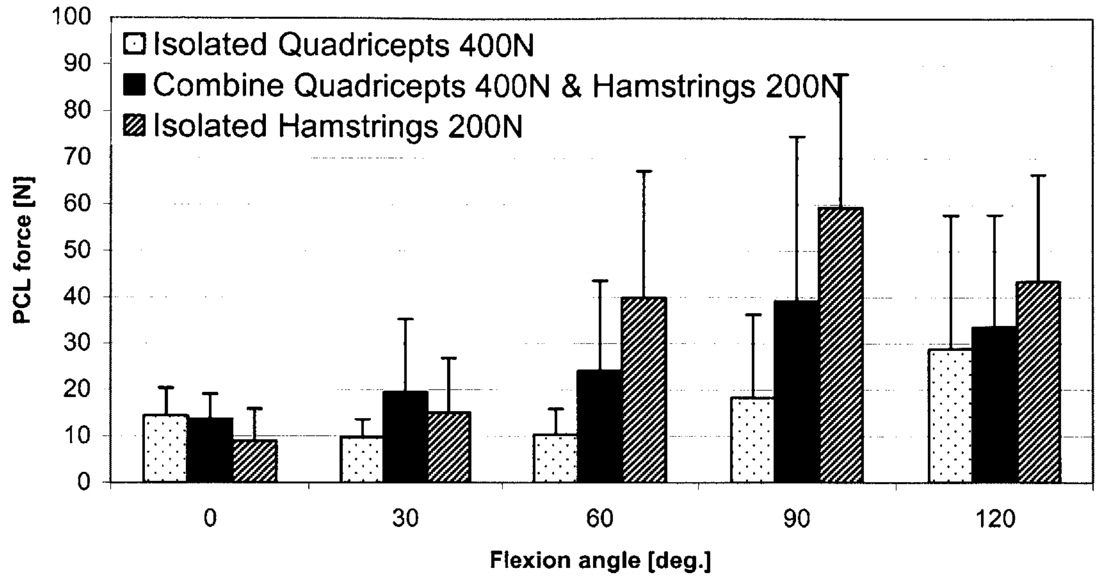


Figure 27: PCL forces in a conventional PCL-retaining TKA as a function on knee flexion under various muscle loads.

The PCL forces in the conventional PCL-retaining TKA were compared to the previously reported PCL forces found in the intact knee (Chapter 4). Under isolated 400N quadriceps load and combined muscle load (400N quadriceps and 200N hamstrings) the PCL forces in the TKA were significantly lower ($p<0.05$) than that measured in the intact knee at 60° and 90° flexion (Figure 28). Under isolated 200N hamstrings load, the PCL forces in the conventional PCL-retaining TKA were significantly lower ($p<0.05$) than that measured in intact knee at 30°, 60° and 90° of flexion.

5.3.3 The Effect of the PCL on Knee Kinematics (femoral translation and tibial rotation) – Study 2

5.3.3.1 Femoral Translation and Tibial Rotation on the Passive Path – Study 2

The unloaded intact knee at full extension served as a reference for all sequential tests. With increasing knee flexion, the center of the intact knee gradually translated posteriorly reaching 8.3 ± 4.3 mm at 60°, 19.2 ± 7.2 mm at 120°, and a peak value of 28.2 ± 9.3 mm at 150° (Figure 29).

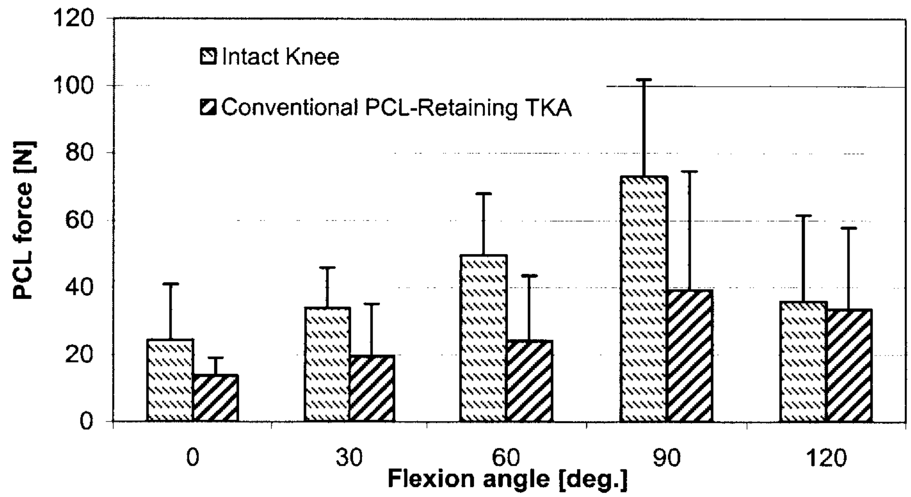


Figure 28: A comparison of PCL forces in the intact knee and conventional PCL-retaining TKA under combined muscle load.

The conventional PCL-retaining TKA knee center was located anteriorly (-2.3 ± 4.0 mm) at full extension compared to the intact knee (Figure 29). It remained in an anterior position for flexion angles less than 60°. At 60° the conventional PCL-retaining knee center was located 3.7 ± 3.6 mm posteriorly compared knee full extension. With further increase in knee flexion, the conventional PCL-retaining knee center continued to translate posteriorly reaching 19.6 ± 5.5 mm at 120°, and a peak value of 23.6 ± 8.7 mm at 150°. At 150°, the conventional PCL-retaining TKA recovered approximately 84% of intact knee translation at the same flexion angle (Figure 29).

The high flexion PCL-retaining TKA closely followed the conventional design (Figure 29). Below 60°, the high flexion PCL-retaining knee center was located anteriorly compared to the unloaded intact knee at full extension. At 60°, the high flexion PCL-retaining TKA was located 1.8 ± 5.9 mm posteriorly compared to the knee at full extension. It continued to translate posteriorly with further increase in knee flexion reaching a value of 16.2 ± 7.3 mm at 120°, and a peak value of 20.6 ± 9.7 mm at 150° which was approximately 73% of intact knee (Figure 29).

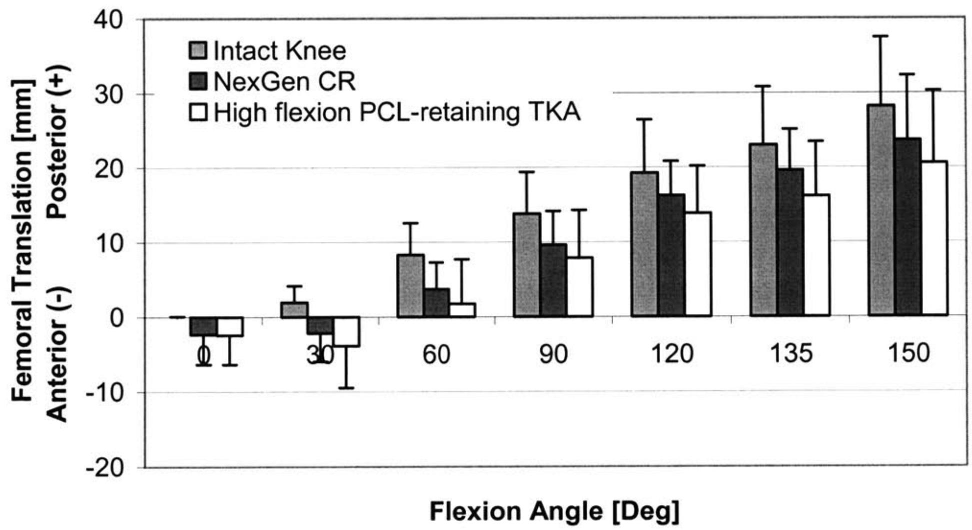


Figure 29: Anterior (-) and posterior (+) femoral translation of the intact, conventional PCL-retaining TKA, and high flexion PCL-retaining TKA on the passive path.

For all knees, internal tibial rotation increased with increasing flexion angle reaching a peak value at 150° (Figure 30). The intact knee reached a maximum value of 9.6° ± 6.3° at 150° flexion. After reconstruction with the conventional PCL-retaining TKA, the tibia rotated internally 1.9° ± 3.8° at full extension and the amount of rotation remained relatively unchanged for flexion angles of less then 120°. Thereafter, internal tibial rotation increased gradually (Figure 30). After reconstruction with the high flexion PCL-retaining TKA, the tibia rotated internally 2.1° ± 3.9° at full extension and the amount of rotation remained relatively unchanged for flexion angles of less than 60°. Thereafter, internal tibial rotation increased gradually with increasing flexion angle reaching a value of 5.4° ± 10.2° at 120° and a peak value of 8.0° ± 7.0° at 150° (Figure 30). There were no statistically significant differences (p>0.05) in terms of the amount of internal tibial rotation between the intact knee and both TKAs throughout the range of passive knee motion.

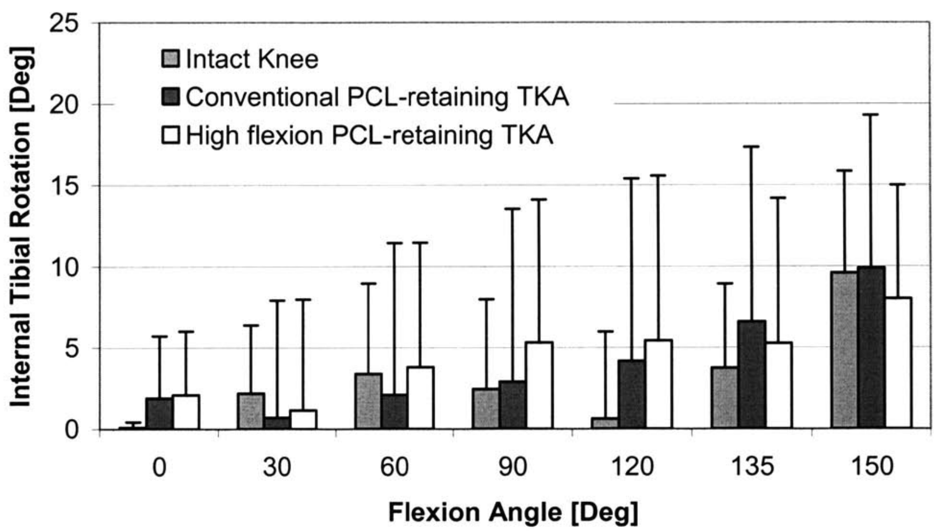


Figure 30: Internal tibial rotation of the intact, conventional PCL-retaining TKA, and high flexion PCL-retaining TKA on the passive path.

5.3.3.2 Femoral Translation under Combined Muscle Load – Study 2

At full extension, the lateral condyle of the intact knee was positioned 4.1 ± 1.6 mm posteriorly when compared to the unloaded intact knee at full extension (Figure 31). Similar behavior was noted for the conventional PCL-retaining TKA design (NexGen) and the high flexion PCL-retaining TKA design (3.3 ± 4.7 mm and 3.8 ± 4.5 mm, respectively). At 30° of flexion, the lateral condyle of the intact knee continued to translate posteriorly reaching a value of 7.0 ± 3.3mm. However, at this flexion angle, the lateral condyle of both TKA designs translated anteriorly reaching a value of -0.4±4.8mm (conventional) and -1.7 ± 5.4mm (high flexion). With increasing flexion angles, the lateral condyle of all knees translated posteriorly reaching a peak value at 150°. The lateral condyle of the intact knee reached a maximum of 32.5 ± 7.8 mm while the lateral condyle of the conventional PCL-retaining design reached a peak of 27.2 ± 8.4 mm (Figure 31). The lateral condyle of the high flexion PCL-retaining design followed conventional TKA closely reaching a maximum value of 24.8 ± 9.0 mm at 150° of knee flexion. At every flexion angle, except at full extension, the lateral femoral condyle translation of both TKA designs was statistically different ($p<0.05$) than that observed in the intact knee. Only at 150°, the translation of the lateral femoral condyle of the high flexion design was statistically different ($p=0.044$) than that of the conventional design.

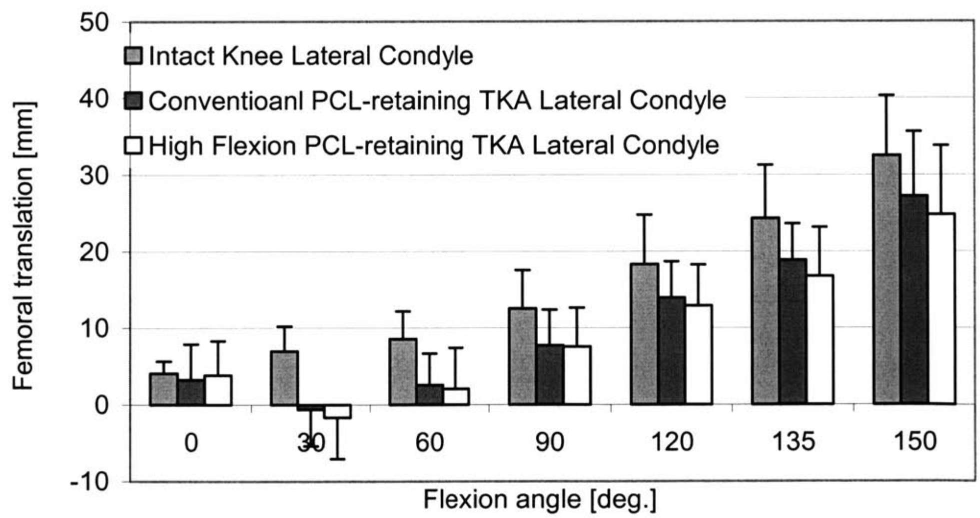


Figure 31: The graph shows the posterior (+) femoral translation of the lateral femoral condyles as a function of knee flexion under combined muscle load (* denotes $p < 0.05$).

Little posterior translation was observed for the medial femoral condyles (0.7 ± 0.8 mm) of the intact knee at full extension (Figure 32). At full extension, the conventional PCL-retaining TKA was positioned slightly posteriorly (0.7 ± 4.4 mm) at the medial condyle when compared to the intact knee passive position at full extension. At the same angle, the medial femoral condyle of the high flexion PCL-retaining TKA was positioned posteriorly at 0.8 ± 4.2 mm. As with the lateral condyle at 30° the medial condyle of the intact knee was positioned posteriorly (2.0 ± 3.0 mm); yet, in both designs, the medial condyles translated anteriorly reaching a minimum value of -1.0 ± 6.0 mm for conventional and -3.0 ± 6.7 mm for high flexion designs. With increasing knee flexion beyond 30°, the medial condyles of all knees translated posteriorly reaching a peak value at 150°. At 150°, the medial condyle of the intact knee was positioned 25.5 ± 9.6 mm posteriorly. At the same flexion angle, the medial condyle of both the conventional design and high flexion design was positioned posteriorly with values of 21.7 ± 12.2 mm and 19.3 ± 11.0 mm, respectively (Figure 32). Except at full extension, the medial condyle femoral translation for both PCL-retaining designs were statistically different ($p<0.05$) than that observed in the intact knee. No statistical significant difference ($p>0.05$) with regard to the medial femoral condyle translation was noted between NexGen CR and CR-Flex TKAs throughout the entire range of flexion.

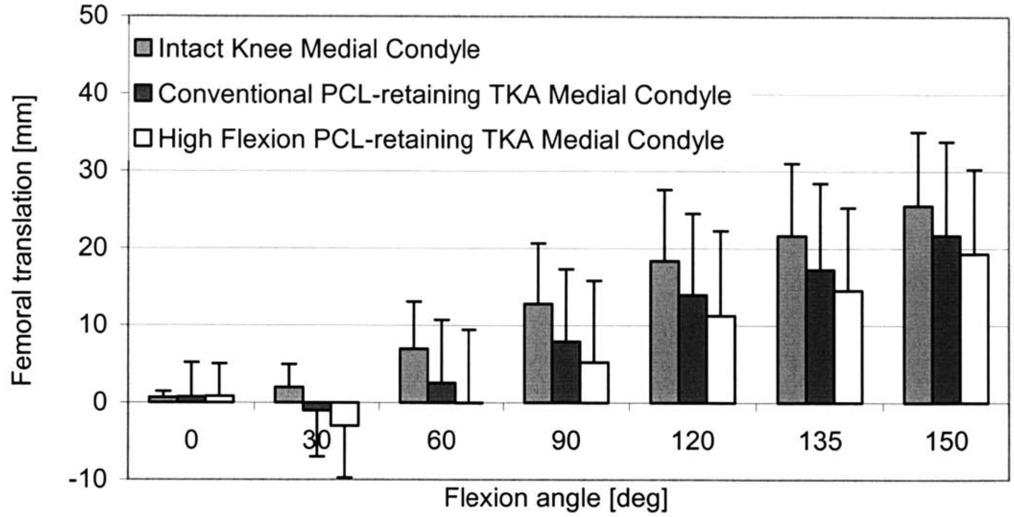


Figure 32: The graph shows the posterior femoral translation of the medial femoral condyles as a function of knee flexion under combined muscle load (* denotes $p < 0.05$).

5.3.3.3 Tibial Rotation under Combined Muscle Load – Study 2

As seen in Figures 31 and 32, the lateral femoral condyles for all knees exhibited a greater posterior femoral translation than the medial femoral condyles signifying the presence of internal tibial rotation. For all knees, internal tibial rotation increased with increasing flexion angles, reaching a maximum value at 150° (Figure 33). No statistical significance was noted between knee states. For example, the tibia of the conventional PCL-retaining TKA internally rotated 1.9° ± 6.7° at 90° and peaked at 150° (6.6° ± 7.3°).

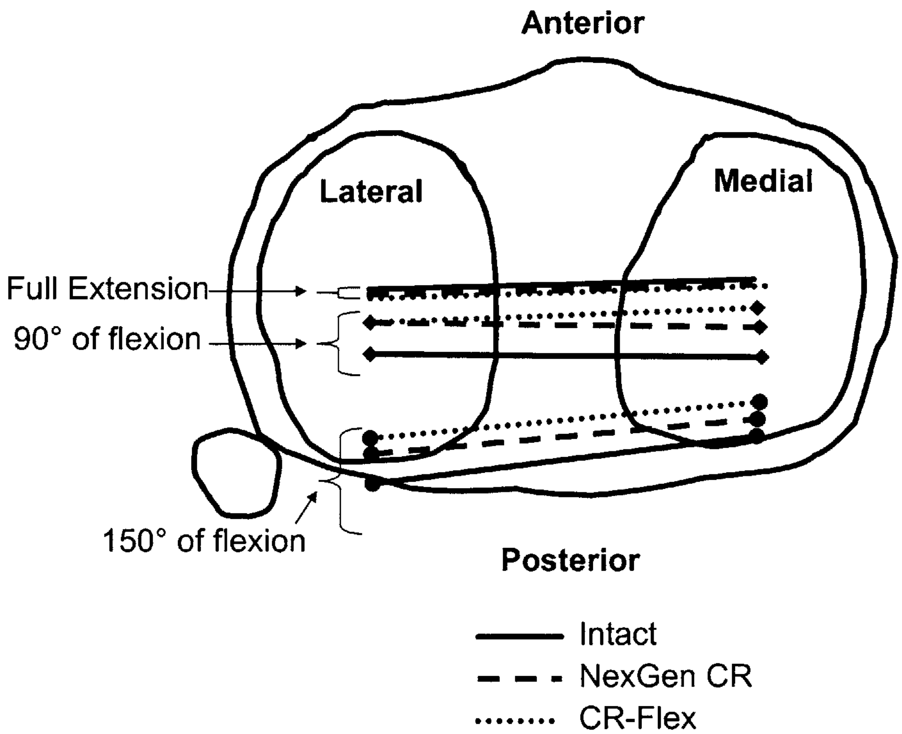


Figure 33: The graph shows the differential motion between the lateral and medial femoral condyles for the three knee states at selected flexion angles under combined muscle load.

5.3.4 The PCL force in a high flexion PCL-Retaining Total Knee Arthroplasty design – Study 2

On the passive path, minimal PCL forces (<30N) were observed throughout the entire range of knee flexion (Figure 34). Under combined muscle load, PCL force in the high flexion PCL-retaining TKA was minimal at full extension (11.9 ± 6.4 N) and at 30° (7.2 ± 4.4 N). With increasing knee flexion, the force in the PCL gradually increased, reaching a peak value of 55.3 ± 29.5 N at 90°. With knee flexion beyond 90°, the PCL force in the PCL-retaining TKA steadily decreased reaching a value of 23.0 ± 12.4N at 150° (Figure 34).

The PCL forces in the high flexion PCL-retaining TKA were compared to the PCL forces recorded in the intact knee and are shown in Figure 35. Under combined muscle load (400N quadriceps and 200N hamstrings), the PCL forces in the high flexion PCL-retaining TKA were significantly lower ($p<0.05$) than that measured in the intact knee at 0° and 30° flexion

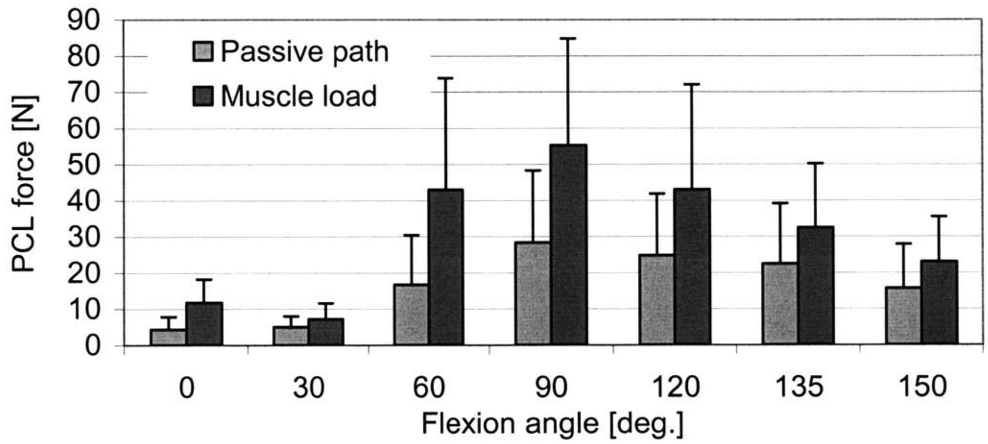


Figure 34: PCL forces in a high flexion PCL-retaining TKA as a function on knee flexion on the passive path and under combined 400N quadriceps and 200N hamstrings load.

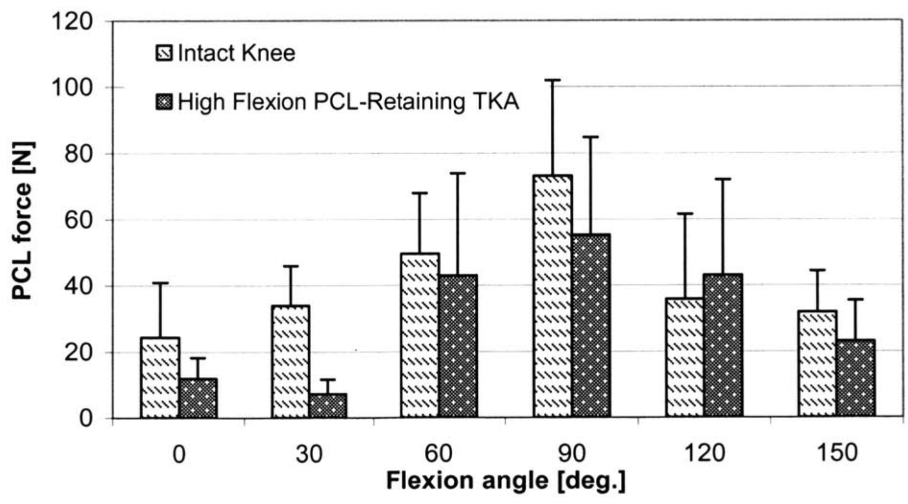


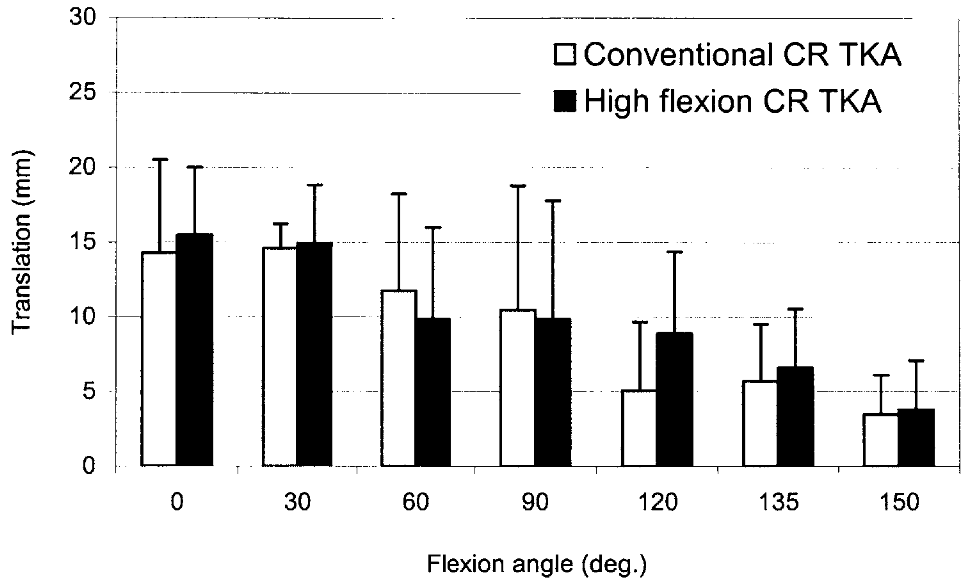
Figure 35: A comparison of PCL forces in the intact knee and the high flexion PCL-retaining TKA under combined muscle load.

5.3.5 Contact Mechanics – Study 2

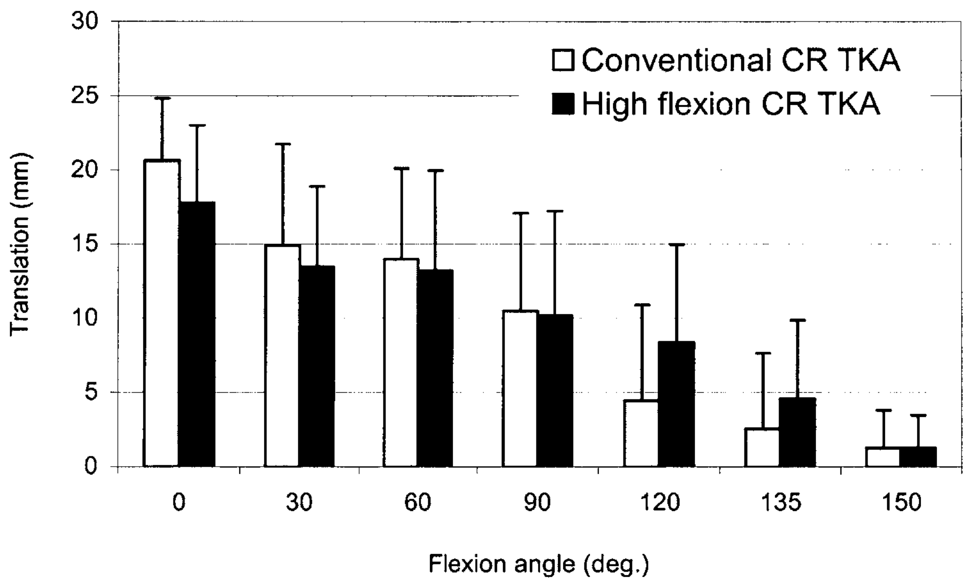
5.3.5.1 Contact Point Location

At full extension, the conventional PCL-retaining femoral component was situated slightly posterior with respect to the mid-coronal plane. At full extension, the medial condyle was positioned 14.3 ± 6.6 mm anteriorly to the posterior edge of the tibial polyethylene (Figure 36A). The lateral condyle was located 20.6 ± 6.3 mm anterior to the posterior edge of the tibial polyethylene (Figure 36B). With increasing flexion, the medial and lateral contact points translated posteriorly. By 90° of flexion, the medial and the lateral condyles had translated posteriorly to a position of approximately 10.5 mm anterior to the posterior edge of the polyethylene. At 150°, the medial and lateral contact points were positioned at 3.8 ± 2.1 mm and 1.3 ± 2.5 mm, respectively, from the posterior polyethylene edge (Figure 36A-B).

The high flexion PCL-retaining TKA exhibited similar behavior with regard to the peak contact point location behavior. At full extension, the high flexion PCL-retaining femoral component was positioned slightly posteriorly to the center of the polyethylene liner at 15.5 ± 4.5 mm (medial) and 17.8 ± 5.2 mm (lateral) from the posterior edge of the tibial polyethylene (Figure 36A-B). Medial and lateral condyles translated posteriorly with increasing flexion reaching a value of 9.8 ± 7.9 mm and 10.2 ± 7.1 mm, respectively, from the posterior edge at 90° of flexion. At 150° of flexion, the medial peak contact pressure point was recorded at 3.1 ± 3.3 mm (Figure 36A) and the peak lateral contact point was 1.0 ± 1.9 mm (Figure 36B) away from the posterior edge of the polyethylene.



A.



B.

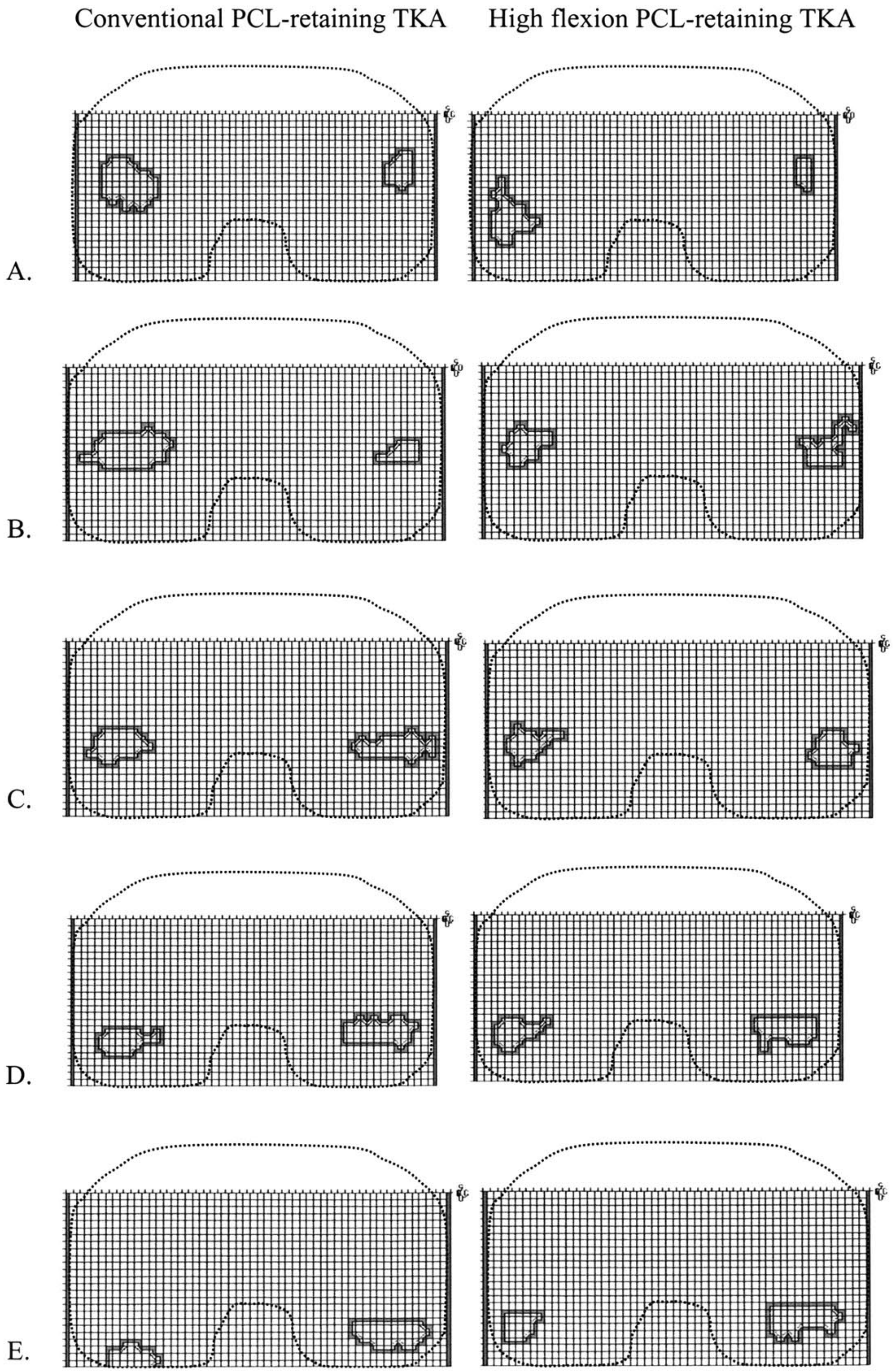
Figure 36: Peak contact point on the A) medial and B) lateral femoral condyles for two TKA designs under combined muscle load (zero Y-axis represents the posterior edge of the polyethylene liner) CR ≡ PCL-retaining

5.3.5.2 Contact Area

Figure 37 A-G illustrates typical contact areas for the conventional and the high flexion PCL-retaining designs as a function of flexion angle. At zero and 90° of flexion, the medial contact area was smaller than the lateral contact area for both the conventional and high flexion PCL-Retaining TKA designs, but the difference was not statistical significant (p>0.05). For example, at 90°, the contact area for the conventional PCL-Retaining component was 22.8 ± 10.3mm^2 for the medial condyle and 27.4 ± 5.0mm^2 for the lateral condyle where as the contact area for the high flexion design was 18.4 ± 13.8mm^2 and 26.3 ± 6.1mm^2 for the medial and lateral condyles, respectively. In addition, no statistically significant difference was found between the two TKA designs with regards to contact area at these data points (Table 2). Inverse relationships were found between flexion and contact area for both TKA designs (i.e. with increasing flexion, the contact area decreased). At 150°, the contact area of the conventional TKA was 13.9 ± 6.9mm^2 and 6.5 ± 8.8mm^2 for the medial and lateral condyles, respectively. At the same flexion angle, the high flexion TKA contact area was 24.5 ± 33.2mm^2 for the medial condyle and 7.3 ± 9.6mm^2 for the lateral condyle (Table 2). In addition to posterior translation of the contact area by 150° of flexion for both TKA designs, the high flexion PCL-retaining TKA exhibited, on average, a larger contact area than that observed for the conventional PCL-retaining with regards to both medial and lateral sides (Figure 38).

Table 2: The average contact area (mm^2) for the two TKAs at selected flexion angles.

NexGen CR		**0°**	**90°**	**150°**
	Medial Condyle	14.9 ± 11.6	22.8 ± 10.3	13.9 ± 6.9
	Lateral Condyle	40.9 ± 20.1	27.4 ± 5.0	6.5 ± 8.8
CR-Flex				
	Medial Condyle	16.3 ± 11.2	18.4 ± 13.8	24.8 ± 33.2
	Lateral Condyle	59.2 ± 61.1	26.3 ± 6.1	7.3 ± 9.6


Conventional PCL-retaining TKA
High flexion PCL-retaining TKA
A.
B.
C.
D.
E.

F.

G.

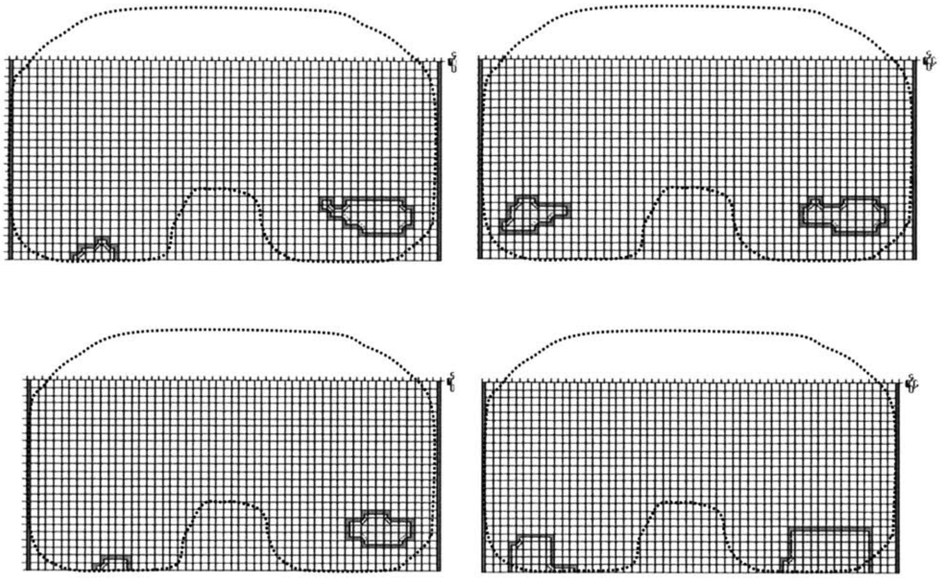


Figure 37: Contact area location of conventional PCL-retaining TKA (left) and high flexion PCL-retaining TKA (right) as a function of knee flexion at A) full extension, B) 30°, C) 60°, D) 90°, E) 120°, F) 135°, and G) 150°.

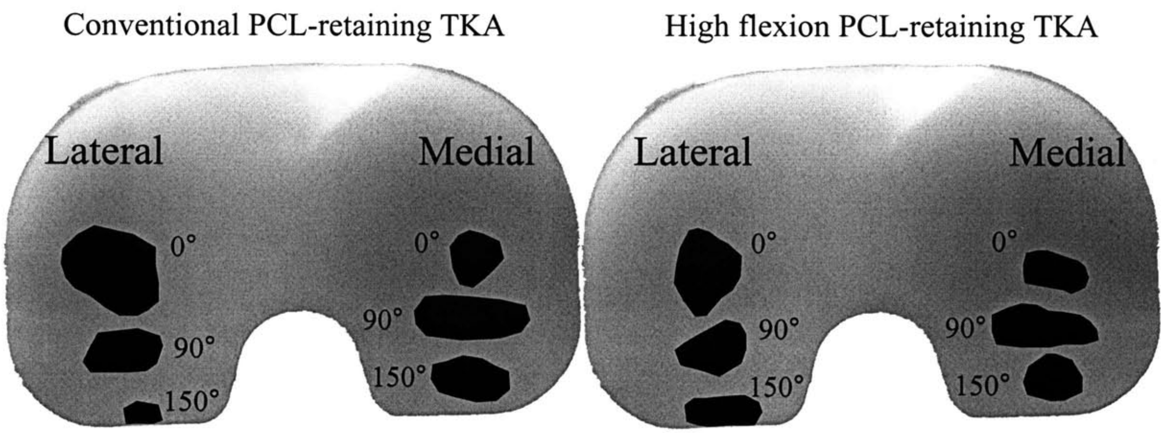


Figure 38: Schematic illustration of the contact area of a typical specimen for a conventional PCL-retaining (NexGen CR) TKA (left) and a high flexion PCL-retaining (CR-Flex) TKA (right) at selected flexion angles under combined muscle load (contact value are in Table 1).

5.4 Conclusions

The PCL is thought to play an essential role in knee joint stability and function after PCL-retaining TKA [1]. However, conflicting data as to the role of the PCL have been reported in the literature [2, 16]. Precise comparisons with normal knees were not possible in these studies owing to their in vivo nature. Furthermore, none of these studies were able to quantify the forces within the PCL after TKA; therefore, implicating the PCL in the kinematic changes after TKA has been challenging. The two studies presented here investigated the effect of the PCL on the kinematics and contact mechanics of the knee after two PCL-retaining reconstructions using a robotic testing system. In both experiments, the intact knee kinematics were used as a baseline to eliminate inter-specimen variations [22].

A reduction in posterior femoral translation after a conventional PCL-retaining TKA was measured when the knee was subjected to the simulated muscle loads. Similar kinematics data have been reported in the literature in both *in vivo* and *in vitro* studies [2, 5, 14, 16]. Mahoney et al [5] measured lower posterior femoral translation after a PCL-retaining TKA (Miller-Galante, Zimmer, Warsaw, IN) in simulated stair climbing. Tarnowski et al [14] reported reduced anterior-posterior translation of a PCL-retaining TKA (NexGen, Zimmer, Warsaw, IN) during gait. Banks et al [16] showed a reduced, but notable posterior femoral translation in PCL-retaining TKA (AMK, DePuy). Dennis et al [2] showed that the femur in PCL-retaining TKA (Press-Fit Condylar Designs, J&J, Warsaw, IN) tended to translate anteriorly in mid flexion during deep knee bends, similar to the ACL deficient knee. This observation is similar to our data on medial femoral condyle that translated anterior in mid flexion. It is difficult to make a quantitative comparison between our study and the data reported in literature due to various factors. These include the different geometrics in the components used, different experimental set-up as well as the difference between our simulated isometric knee flexion-extension experiments and theirs which involved deep knee-bending [2], gait [14], and step-up [16]. In our experiment, the intact knee and the TKAs used the same reference (knee coordinate system) to measure the knee joint motion, while *in vivo* studies usually compared a group of normal subjects with another group of TKA patients. The variability between normal subjects and TKA patients may be difficult to eliminate.

The posterior femoral translation of both PCL-retaining TKA designs was found to be influenced by the forces in the PCL. The *in situ* forces of the PCL in the PCL-retaining TKA increased with increasing flexion up to 90° but decreased beyond 90° (Figure 39). These trends are consistent with those measured in normal knees. However, the magnitude of the PCL forces in normal knees was greater than that measured in either PCL-retaining TKA. A peak force of 73 N was measured in the PCL of intact knees at 90° flexion under the combined muscle load, while the corresponding force in the PCL after conventional PCL-retaining TKA was only 38 N and after a high flexion PCL-retaining TKA was 55 N. This lower PCL force may be due to various factors. Insall [3] discussed the difficulty in achieving the correct tension in the PCL during TKA. If the PCL is not tight, then either the balance is perfect, or the PCL is too loose. Currently, there is no reliable way to assess a PCL that is too loose. In the case of a PCL that is too tight, an incremental release has been recommended [8, 31] until soft tissue balance is achieved. This release reduces the overall stiffness of the PCL, resulting in a weaker ligament and an increased flexion gap. Interestingly, other investigators have found some non-functioning posterior cruciate ligaments in patients who have had a PCL-retaining TKA [15]. These authors argued that this so called "functioning PCL group" might, in fact, represent patients in whom the PCL is not the sole contributor to posterior stability.

The role of the PCL as the primary constraint in posterior knee stability may therefore be reduced.

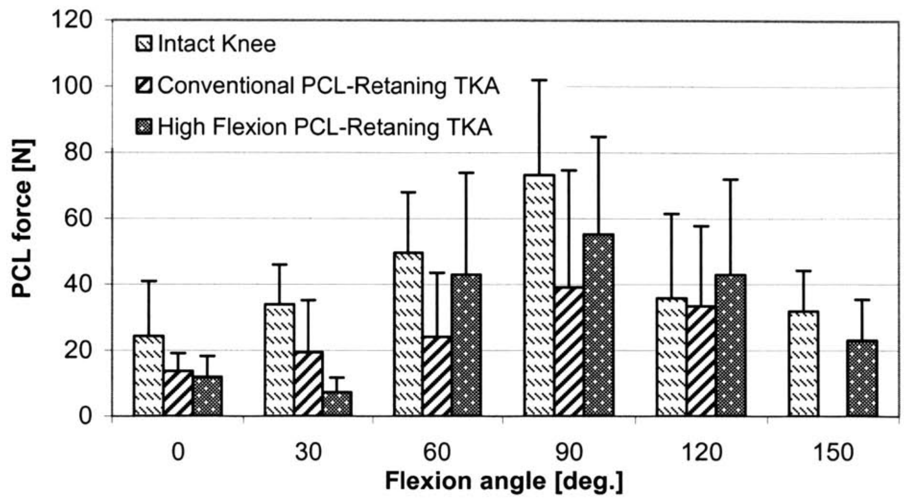


Figure 39: A comparison of PCL forces in all three knee states under combined muscle load. Note that the conventional TKA was not tested at 150°.

Even though our data demonstrated that the PCL of the TKA might not be loaded to the level of the intact knee, the data demonstrated that the PCL *did* contribute to posterior femoral translation by the fact that posterior femoral translation was further reduced in the TKA when the PCL was transected. The difference between the femoral translation of the TKA and the intact knee was not as marked when the ligament was retained compared to that after PCL transection. We conclude that the PCL is an important structure in improving posterior femoral translation after PCL-retaining TKA.

Reduced posterior femoral translation has important clinical implications for the patellofemoral joint. It has been observed that failure of the patellofemoral articulation is a common cause of failure and revision of TKA [32-34]. Patellofemoral contact force is, in part, dependent on the position of the tibial tuberosity in the anterior-posterior direction [35-37]. In the PCL-retaining TKA, a relatively posteriorly translated tibia may effectively result in increased patellofemoral contact forces. Reduced posterior femoral translation throughout the range of flexion will also result in a smaller lever arm for the patellar tendon, thus decreasing the efficiency of the extensor mechanism of the knee. Such changes may contribute to the complications seen in the patellofemoral joint after TKA. Therefore, reducing contact stresses at this articulation is of great importance.

The internal tibial rotation of the intact knee measured in these studies was similar to that reported in the literature [14, 22, 38-40]. The presence of the PCL in the TKA did not alter the rotational behavior of the knee. In both the PCL-retaining and PCL-deficient TKAs, internal tibial rotation increased with increasing flexion. Internal tibial rotation under muscle loads was substantially reduced at 30° of flexion in both the conventional PCL-retaining and PCL-deficient TKAs compared to that observed in the intact knee. Importantly, the tibial rotation of the TKAs was not significantly different from that of the intact knee beyond 30° of flexion. This may be due to the asymmetry between the sagittal radii of the medial and lateral condyles of the femoral prosthesis. The radius of the lateral condyle is greater than the medial condyle, resulting in internal tibial rotation. This effect is seen regardless of whether the PCL is intact or resected.

The reduced internal tibial rotation at 30° after a PCL-retaining TKA under the simulated muscle loads may have important clinical implications. Most daily functions require knee flexion around 30°. Reduced internal tibial rotation may increase the inclination of the patella tendon with respect to the tibial axis. This would produce a larger lateral component of the quadriceps force. Consequently, the lateral side of the patella would be subjected to excessive contact pressures, which has been shown in recent studies [30]. In combination with reduced posterior femoral translation, reduced internal tibial rotation would be more likely to accelerate patellofemoral joint complications, leading to revision of the arthroplasty.

The contact area and the peak contact pressure point location in high flexion PCL-retaining TKA and a conventional PCL-retaining TKA were also examined. The former incorporates elongated femoral condyles compared to the conventional knee. This design feature is intended to prevent edge loading and to facilitate larger contact areas at high knee flexion. We found that both TKA designs exhibited similar posterior femoral translation; yet the contact area of the high flexion PCL-retaining design at high flexion was larger than that observed in the conventional PCL-retaining component. Despite the marked difference observed between the two designs with regard to contact area, based on the number of specimens in this particular study, this difference was not found to be statistically significant ($p>0.05$).

The femorotibial contact positions for two TKA designs (posterior stabilized and CR TKAs; Press-Fit Condylar, DePuy Johnson & Johnson, Warsaw, IN) have been studied in an in-vivo analysis [41-43]. Using two-dimensional radiographic images and a fitting technique, Komistek et al [41] reported that for a sitting position, the medial and lateral condyle for both TKA designs were positioned posteriorly with respect to the mid-coronal plane. This is similar to the findings reported in this study. We found that at full extension, both components were positioned slightly posteriorly to the center of the polyethylene.

An important factor that may affect the outcome of any TKA at high flexion angles is the loading characteristics of the knee beyond 120°. Nagura et al [44] reported that at high flexion angles, large posterior forces (58.3-67.8% body weight) and large moments were recorded for deep flexion activities. These large moments and forces will result in high stress at high knee flexion. For a given contact area, an increase in a joint force directly correlate with an increase in the join stress, increasing polyethylene peak contact pressure, thus jeopardizing the longevity of the implant. However, for a given force, a larger contact area would yield a reverse phenomenon, reducing the contact stress and pressure and extending the life span of the implant. Our study showed that the high flexion PCL-retaining TKA provided a larger contact area on both the medial and lateral condyles at high knee flexion.

The data in this study revealed that, on average, the peak contact point location for the conventional PCL-retaining design was situated more posteriorly than that of the high flexion PCL-retaining design at flexion angles greater that 90°. Point loading of the femoral condyles with the polyethylene edge for the conventional PCL-retaining TKA was observed at 120°, 135° and 150°. At 150°, the femoral condyle was no longer articulating with the polyethylene but rather articulated on the bone. This extreme posterior contact with the tibial edge may result in high pressure at the edge and may lead to delamination and/or deformation of the polyethylene. Furthermore, posterior edge-loading may result in internal instability of the knee, and this may represent a significant biomechanical factor hindering high flexion in patients following TKA. The tibial component should provide for enough posterior coverage in order to effectively elongate the articular surface and reduce the potential for edge-loading, improving knee stability at high flexion. Both consequences will affect the outcome and the longevity of an implant. One solution is to increase stability as a result of a more consistent contact area, as the high flexion PCL-retaining TKA was designed for. The potential instability with regards to posterior contact in combination with the calculated loads at high flexion, can lead to severe high stress at the posterior edge of the poly and ultimately loss of contact with the posterior polyethylene in extreme conditions. In this study, the high flexion PCL-retaining design did not reach the polyethylene edge until 150° which was approximately 20° more than for the conventional design.

In conclusion, we investigated the kinematics, PCL force, and contact mechanics of two TKA designs (conventional and high flexion PCL-retaining TKA) using a robotic testing system. We concluded that kinematics (posterior femoral translation and internal tibial rotation) are essential in achieving high knee flexion but it is not sufficient. The bell-curved-shape PCL force demonstrated that the PCL contributes to posterior femoral translation for all knees and its largest contribution is at the mid flexion range. The decrease in the forces in the PCL after TKA correlated with altered knee kinematics. We have observed that below 120°, an increase in PCL force coincided with an increase in posterior femoral translation. However, at high knee flexion, despite the low PCL force in the high flexion PCL-retaining TKA, posterior femoral translation continued to rise and is fact, it restored 80% of intact knee kinematics. The enhanced posterior femoral translation beyond 120° may be caused by the compression of posterior soft tissue, including posterior capsule, hamstring muscles, skin, and fat between the tibial and femoral shafts. Further posterior femoral translation becomes possible at high flexion as the compressed soft tissue pushes the tibia anteriorly and the femur posteriorly. However, with an increase in posterior femoral translation, there is a tendency toward edge loading which limits further flexion. In view of the fact that the TKA may not completely restore this soft tissue compression, edge loading after TKA becomes more dominant in affecting knee stability and the ability of patients to flex to higher degrees of flexion. The information in this study provides surgeons and prostheses designers with a deeper insight into the role of the PCL in TKA, and potential ways to improve the function of PCL-retaining TKA.

5.5 References

1. Andriacchi, T. and J. Galante, *Retention of the posterior cruciate in total knee arhtroplasty.* J Arthroplasty, 1988. Supplement: p. S13-S19.
2. Dennis, D.A., et al., *In vivo knee kinematics derived using an inverse perspective technique.* Clin Orthop, 1996(331): p. 107-17.
3. Insall, J.N., *Presidential address to The Knee Society. Choices and compromises in total knee arthroplasty.* Clin Orthop, 1988(226): p. 43-8.
4. Laskin, R.S. and H.M. O'Flynn, *The Insall Award. Total knee replacement with posterior cruciate ligament retention in rheumatoid arthritis. Problems and complications.* Clin Orthop, 1997(345): p. 24-8.
5. Mahoney, O.M., et al., *Posterior cruciate function following total knee arthroplasty. A biomechanical study.* J Arthroplasty, 1994. 9(6): p. 569-78.
6. Pereira, D.S., F.F. Jaffe, and C. Ortiguera, *Posterior cruciate ligament-sparing versus posterior cruciate ligament-sacrificing arthroplasty. Functional results using the same prosthesis.* J Arthroplasty, 1998. 13(2): p. 138-44.
7. Singerman, R., et al., *Decreased posterior tibial slope increases strain in the posterior cruciate ligament following total knee arthroplasty.* J Arthroplasty, 1996. 11(1): p. 99-103.
8. Worland, R.L., D.E. Jessup, and J. Johnson, *Posterior cruciate recession in total knee arthroplasty.* J Arthroplasty, 1997. 12(1): p. 70-3.

9. Andriacchi, T.P., J.O. Galante, and R.W. Fermier, *The influence of total knee-replacement design on walking and stair-climbing.* J Bone Joint Surg Am, 1982. 64(9): p. 1328-35.

10. Lewandowski, P.J., et al., *Kinematics of posterior cruciate ligament-retaining and -sacrificing mobile bearing total knee arthroplasties. An in vitro comparison of the New Jersey LCS meniscal bearing and rotating platform prostheses.* J Arthroplasty, 1997. 12(7): p. 777-84.

11. Freeman, M.A. and G.T. Railton, *Should the posterior cruciate ligament be retained or resected in condylar nonmeniscal knee arthroplasty? The case for resection.* J Arthroplasty, 1988. 3 Suppl: p. S3-12.

12. Ritter, M., et al., *Long-term survival analysis of a posterior cruciate-retaining total condylar total knee arthroplasty.* Clin Ortho, 1994. 309: p. 136-145.

13. Arima, J., et al., *Effect of partial release of the posterior cruciate ligament in total knee arthroplasty.* Clin Orthop, 1998(353): p. 194-202.

14. Tarnowski, L.E., et al., *Three-dimensional motion of cruciate retaining and posterior stabilized TKAs during walking.* Trans ORS, 1998. 44: p. 804.

15. Matsuda, S., et al., *Knee stability in posterior cruciate ligament retaining total knee arthroplasty.* Clin Orthop, 1999(366): p. 169-73.

16. Banks, S.A., G.D. Markovich, and W.A. Hodge, *In vivo kinematics of cruciate-retaining and -substituting knee arthroplasties.* J Arthroplasty, 1997. 12(3): p. 297-304.

17. Stiehl, J.B., et al., *Fluoroscopic analysis of kinematics after posterior-cruciate-retaining knee arthroplasty.* J Bone Joint Surg Br, 1995. 77(6): p. 884-9.

18. Stiehl, J.B., R.D. Komistek, and D.A. Dennis, *Detrimental kinematics of a flat on flat total condylar knee arthroplasty.* Clin Orthop, 1999(365): p. 139-48.

19. Kim, H., et al., *Rollback in posterior cruciate ligament-retaining total knee arthroplasty. A radiographic analysis.* J Arthroplasty, 1997. 12(5): p. 553-61.

20. Dennis, D.A., et al., *Range of motion after total knee arthroplasty: the effect of implant design and weight-bearing conditions.* J Arthroplasty, 1998. 13(7): p. 748-52.

21. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

22. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.

23. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

24. Li, G., et al., *The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL.* J Biomech, 1999. 32(4): p. 395-400.

25. Li, G., et al., *Prediction of antagonistic muscle forces using inverse dynamic optimization during flexion/extension of the knee.* J Biomech Eng, 1999. 121(3): p. 316-22.

26. Hoher, J., et al., *In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study.* J Orthop Res, 1999. 17(5): p. 763-8.

27. Grood, E.S. and W.J. Suntay, *A joint coordinate system for the clinical description of three-dimensional motions: application to the knee.* J Biomech Eng, 1983. 105(2): p. 136-44.

28. Most, E., et al., *Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty.* Clin Orthop, 2003(410): p. 101-13.

29. Fox, R.J., et al., *Determination of the in situ forces in the human posterior cruciate ligament using robotic technology. A cadaveric study.* Am J Sports Med, 1998. 26(3): p. 395-401.

30. Li, G., et al., *Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads--an in vitro experimental study.* J Orthop Res, 2002. 20(4): p. 887-92.

31. Ritter, M.A., P.M. Faris, and E.M. Keating, *Posterior cruciate ligament balancing during total knee arthroplasty.* J Arthroplasty, 1988. 3(4): p. 323-6.

32. Bayley, J.C., et al., *Failure of the metal-backed patellar component after total knee replacement.* J Bone Joint Surg Am, 1988. 70(5): p. 668-74.

33. Collier, J.P., et al., *All-polyethylene patellar components are not the answer.* Clin Orthop, 1991(273): p. 198-203.

34. Lombardi, A.V., Jr., et al., *Fracture/dissociation of the polyethylene in metal-backed patellar components in total knee arthroplasty.* J Bone Joint Surg Am, 1988. 70(5): p. 675-9.

35. Maquet, P., *Advancement of the tibial tuberosity.* Clin Orthop, 1976(115): p. 225-30.

36. Maquet, P., *Mechanics and osteoarthritis of the patellofemoral joint.* Clin Orthop, 1979(144): p. 70-3.

37. Miller, R.K., et al., *In vitro measurement of patellofemoral force after three types of knee replacement.* J Bone Joint Surg Br, 1998. 80(5): p. 900-6.

38. Hirokawa, S., et al., *Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction.* Am J Sports Med, 1992. 20(3): p. 299-306.

39. Lafortune, M.A., et al., *Three-dimensional kinematics of the human knee during walking.* J Biomech, 1992. 25(4): p. 347-57.

40. Iwaki, H., V. Pinskerova, and M.A. Freeman, *Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee.* J Bone Joint Surg Br, 2000. 82(8): p. 1189-95.

41. Komistek, R.D., et al., *In vivo comparison of femorotibial contact positions for press-fit posterior stabilized and posterior cruciate-retaining total knee arthroplasties.* J Arthroplasty, 2002. 17(2): p. 209-16.

42. Dennis, D.A., et al., *In vivo anteroposterior femorotibial translation of total knee arthroplasty: a multicenter analysis.* Clin Orthop, 1998(356): p. 47-57.

43. Dennis, D.A., R.D. Komistek, and M.R. Mahfouz, *In vivo fluoroscopic analysis of fixed-bearing total knee replacements.* Clin Orthop, 2003(410): p. 114-30.

44. Nagura, T., et al., *Mechanical loads at the knee joint during deep flexion.* J Orthop Res, 2002. 20(4): p. 881-6.

Chapter 6

THE ROLE OF THE CAM-SPINE IN POSTERIOR-STABILIZED TOTAL KNEE ARTHROPLASTY

6.1 Introduction

Posterior-stabilized total knee arthroplasty (TKA) was introduced to prevent posterior subluxation of the tibia and to improve knee flexion by promoting femoral rollback (posterior femoral translation) [1, 2]. This is achieved by contact of the femoral cam with the tibial spine of the prosthesis, which prevents excessive posterior tibial translation during knee flexion, promotes posterior femoral translation, and possibly, increases the range of knee flexion after TKA [2]. Good long-term clinical results have been reported in relation to function and prosthesis longevity [3-5]. Understanding the mechanism by which the cam-spine mechanism acts and the subsequent effects on knee kinematics is necessary for further improvements in posterior-stabilized TKA design.

Few studies investigated the biomechanical characteristics of this cam-spine system interaction. In a multi-center study using the Press Fit Condylar design (Johnson & Johnson, Raynham, MA), Dennis et al [6] reported that the posterior-stabilized TKA demonstrated tibiofemoral rollback during in vivo squatting activity of patients but was less than that observed in normal knees. Another study assessed five different knee prostheses [7] and had similar observations as Dennis et al. However, Banks et al [8] found that axial rotation and condylar translations decreased with a posterior-substituting TKA (Primary Posterior Stabilized TKA, Osteonics) during step-up activity. The kinematics of the knee in these studies were indirectly measured by matching the sagittal view pattern of the three-dimensional prosthesis to that of two-dimensional fluoroscopic pictures of the prosthesis obtained during in vivo activities [9]. Tarnowski et al [10] measured the in vivo three-dimensional motion of TKAs (NexGen, Zimmer, Warsaw, IN) during walking using a newly-developed point-cluster technique [11]. They reported that the range of anterior-posterior displacement was significantly reduced for patients with posterior-stabilized TKAs compared with those measured in normal subjects. A direct comparison between the different studies is difficult as each study uses a different set-up and different loading conditions.

Computerized models have also been utilized to conduct parametric studies of TKA kinematics [12, 13]. For example, Delp et al [14] used two-dimensional computer models to examine the effect of position and height of the tibial spine on the tibiofemoral rollback of the knee. They reported that an increase in the spine height had little effect on femoral rollback. However, anterior placement of the tibial spine of the femoral cam reduced femoral rollback. Piazza et al [15] investigated the effect of posterior tilt of the tibial component on the motion of a posterior-stabilized TKA using computer simulation. They reported that femoral rollback (posterior translation) was enhanced by a cam-spine mechanism at higher flexion angles. Any amount of posterior tilt reduced the amount of rollback by limiting the interaction between the cam and spine. For example, they reported that tilting the component 5° posteriorly caused the femoral cam to contact with the tibial spine 18° more than with the un-tilted component.

In all these experimental and computational studies, the native knee kinematics were unknown, therefore making it difficult to objectively evaluate the capability of the TKA in restoring normal knee kinematics. Little quantitative data has been reported on when and how the cam-spine mechanism is effective during flexion-extension of the knee. This information would be invaluable for further improvement of TKA design as well as surgical manipulation of the TKA in order to achieve high flexion of the knee (beyond 120°) after TKA.

Therefore, the objective of this chapter is to provide the reader with information regarding a number of currently available posterior-stabilized TKA designs and discuss the factors influencing knee flexion.

Study 1: In this study, a conventional posterior-stabilized TKA (NexGen LPS, Zimmer, Warsaw, IN) was tested in the flexion range of 0° to 120°. Knee kinematics and cam-spine contact forces in response to simulated muscle loads were measured by employing the robotic experimental set-up. Each specimen was tested in its native state followed by a posterior-substituting TKA thereby limiting inter-specimen variations.

Study 2: A single design posterior-stabilized TKA design offering both fixed- and mobile-bearing tibial components (Zimmer Inc, Warsaw, IN) was tested also using the robotic experimental set-up. The effect of design on knee kinematics (posterior femoral translation and tibial rotation) was examined.

6.2 Study 1: Conventional Posterior-Stabilized TKA (0°-120° of Flexion)

6.2.1 Introduction

It is believed that the contact of the femoral cam with the tibial spine of the posterior-stabilized TKA is necessary for posterior femoral translation. Posterior cruciate retaining TKA (Chapter 5) was introduced to maintain posterior femoral translation by retaining the posterior cruciate ligament. In a posterior-substituting knee replacement, component design must compensate for the deficient posterior cruciate ligament. Femoral rollback in a posterior-stabilized TKA is achieved by the interaction of the femoral cam with the tibial spine [15]. However, little is known about when and how this cam-spine interaction in a posterior-stabilized TKA affects knee range of motion and knee stability. As described above, few biomechanical studies have been performed to investigate the mechanisms of cam-spine interaction and the kinematics of the posterior-stabilized TKA when compared to the normal knee.

This study investigated the mechanism of cam-spine contact in a single design, posterior-stabilized TKA (NexGen LPS, Zimmer Inc., Warsaw, IN) using cadaveric human knee specimens and a robotic testing system. The effect of the cam-spine engagement on knee kinematics was explored and the cam-spine contact forces in response to simulated muscle loads were measured, which were never quantified before.

Each specimen was tested in its native state and followed by a conventional posterior-substituting TKA. Therefore, the inter-specimen variations could be minimized.

6.2.2 Material and Methods

Nine fresh frozen human cadaveric knee specimens (age range 66 ± 6 years) were used in this study. The specimen preparation and mounting on the robotic testing system were described in Sections 4.3.1 and 4.3.2.

6.2.2.1 Passive Path Determination – Intact Knee

The knee was initially pre-conditioned by flexing it between full extension and full flexion ten times manually. After it was mounted onto the joint test system, a series of passive positions were defined during knee flexion/extension between 0° and 120°. This was done using force-moment control [16, 17] of the test system at one degree increments of knee flexion. At each flexion angle, knee positions at the other five degrees-of-freedom were determined such that the residual forces and moments at the intact knee joint center (midpoint of the transepicondylar line) were minimal (below 5 N and 0.5 N-m for normal knees, respectively). This method has a high level of repeatability for finding the passive positions for the native knee and the TKAs [16]. These pre-determined passive positions represented the relative position of the tibia with respect to the femur when the joint carried minimal load. The path was then used as the reference positions for application of simulated physiological loads during subsequent tests.

6.2.2.2 Kinematics Determination – Intact Knee

Each knee was tested at selected flexion angles of 0°, 30°, 60°, 90° and 120° on its passive path in response to muscle loads. Quadriceps and hamstring forces were applied to the joint to simulate knee flexion/extension. For the application of quadriceps muscle loads, the rectus femoris tendon was sutured to a cable and then connected to a weight. For the application of hamstring muscle loads, the medial hamstrings (semimembranosus and semitendinosus) and lateral hamstrings (biceps femoris tendon) were sutured to individual cables and connected to weights. Both the quadriceps and hamstring muscle loads were applied along the direction of the femoral shaft [18]. It has been reported that the quadriceps forces may reach several times body weight during functional activities [19]. However, due to the limitation of the testing system, the three muscle loads chosen were lower than physiological levels. The loads were: 1) isolated quadriceps force of 400 N; 2) combined quadriceps and hamstrings load (400 N/200 N); 3) isolated hamstring force of 200 N. With the knee in the full extension position on its passive path, a muscle load was applied to the joint. The tibia moved in the remaining five degrees-of-freedom until it reached an equilibrium position where the applied load was balanced by the constraint forces generated inside the knee joint. The new knee position was recorded by the robotic manipulator (U_{e1}). The test was repeated at each selected flexion angle (0°, 30°, 60°, 90° and 120°) and for all external loads.

6.2.2.3 Measurement of TKA Kinematics

After measurement of the native knee kinematics in response to muscle loads, the same knee was reconstructed by an orthopaedic surgeon using a conventional posterior-stabilized TKA (NexGen LPS, Zimmer Inc., Warsaw, IN). This component has symmetric radii in medial and lateral condyles. The femoral component has a large distal radius and smaller posterior radius. The contact point of the cam-spine system, in this design, moves down the spine after engagement.

Each knee was radiographed prior to testing and the estimated component size was determined. The posterior-stabilized TKA was inserted via a standard medial parapatellar arthrotomy. The femoral cuts were performed using the intramedullary guide for varus-valgus alignment, and the epicondylar axis for rotational alignment. The tibial cut was made using the extramedullary system, using the tibial crest and center of the tibial plateau as reference points. A cut with a 7° posterior slope was performed using the cutting guides as recommended by the manufacturer, with the component ultimately aligned with the junction of the medial and middle thirds of the tibial tuberosity. The anterior and posterior cruciate ligaments were resected. The patella was not resurfaced. Trial components were inserted and knee stability, passive range of motion, patella tracking and flexion-extension gaps were evaluated. The flexion-extension gaps were equalized by incremental subperiosteal soft tissue releases. The arthrotomy was closed with a continuous suture, as was the skin. All component sizes were available at each experiment, so at no stage did the surgeon had to alter the cuts to fit a particular size. Definitive sizes were determined intra operatively.

The reconstructed knee was tested using identical procedures as described above for the intact knee. A passive path for the reconstructed knee was determined first. The passive path served as a reference for application of external loads, and for measurement of knee kinematics in response to external loads. The muscle loads were applied, the corresponding knee joint kinematics were measured by the robot (U_{e2}) and the constraint forces at the knee center ($\vec{f}_1$) were obtained through the measurement of the six degrees of freedom load cell.

6.2.2.4 Measurement of Cam-Spine Contact Forced

To measure the in-situ contact forces of the femoral cam and the tibial spine, the tibial spine was cut and the above kinematics (U_{e2}) was then replayed on the specimen. The load cell measured the new joint constraint forces and noted by $\vec{f}_2$. By the principle of superposition, the differences between forces measured before and after cutting the tibial spine represent the in-situ contact forces of the femoral cam and tibial spine [17, 18], as shown in the equation below:

$$\vec{f}_{cam-spine} = \vec{f}_1 - \vec{f}_2.$$

6.2.2.5 Statistical Analysis

A two-way repeated measure Analysis of Variance (ANOVA) was used to detect whether the knee state (intact and reconstructed) has a significant effect on knee joint motion. A one-way repeated measure ANOVA was used to detect whether muscle load had a significant effect on the cam-spine force. Significant difference was set to a p value of less than 0.05.

6.2.3 Results

6.2.3.1 Posterior Translation of the Lateral and Medial Femoral Condyles

The posterior translations of lateral and medial femoral condyles of the knee after the TKA were altered in comparison to those of the native knee at all flexion angles (Figure 40). During passive flexion of the knee (under no external load), the lateral femoral condyle of the native knee translated posteriorly from 4.3 ± 3.0 mm (mean ± S.D) at 30° to 18.4 ± 5.0 mm at 120° (Figure 40). After TKA, the lateral femoral condyle moved from 0.0 ± 7.2 mm at 30° to 14.3 ± 5.0 mm at 120°, which was significantly lower ($p<0.05$) than that of the intact knee. On the medial side of the knee, the femoral condyle of the native knee moved posteriorly from 0.1 ± 3.8 mm at 30° to 11.4 ± 11.4 mm at 120° of flexion, while after TKA, it moved from –2.7 ± 6.7 mm at 30° to 6.7 ± 11.5 mm at 120° of flexion. The posterior femoral translation of the medial condyle was significantly reduced after the posterior stabilized TKA beyond 30° of flexion.

Under the quadriceps load, the lateral condyle of the native knee moved posteriorly from 13.1 ± 3.6 mm at 30° to 17.3 ± 5.7 mm at 120° (Figure 41). Following the posterior-stabilized TKA, the lateral condyle translated by 5.5 ± 6.6 mm at 30° and 13.9 ± 5.2 mm at 120°, which were significantly lower (58% and 20% lower) than those of intact native knee. On the medial side, the femoral condyle of the native knee moved posteriorly from 3.0 ± 2.6 mm at 30° to 10.5 ± 11.7 mm at 120° (Figure 41). After the posterior-stabilized TKA, the medial femoral condyle translated posteriorly from 1.2 ± 8.1 mm at 30° to 6.5 ± 11.6 mm at 120°, which was significantly lower than in the intact knee beyond 30° of flexion.

Similar comparisons can be made for the femoral condyle motions under the combined quadriceps-hamstring load and the isolated hamstring load. The posterior femoral condyle motion was significantly reduced beyond 30° of flexion after TKA (Figures 42 and 43).

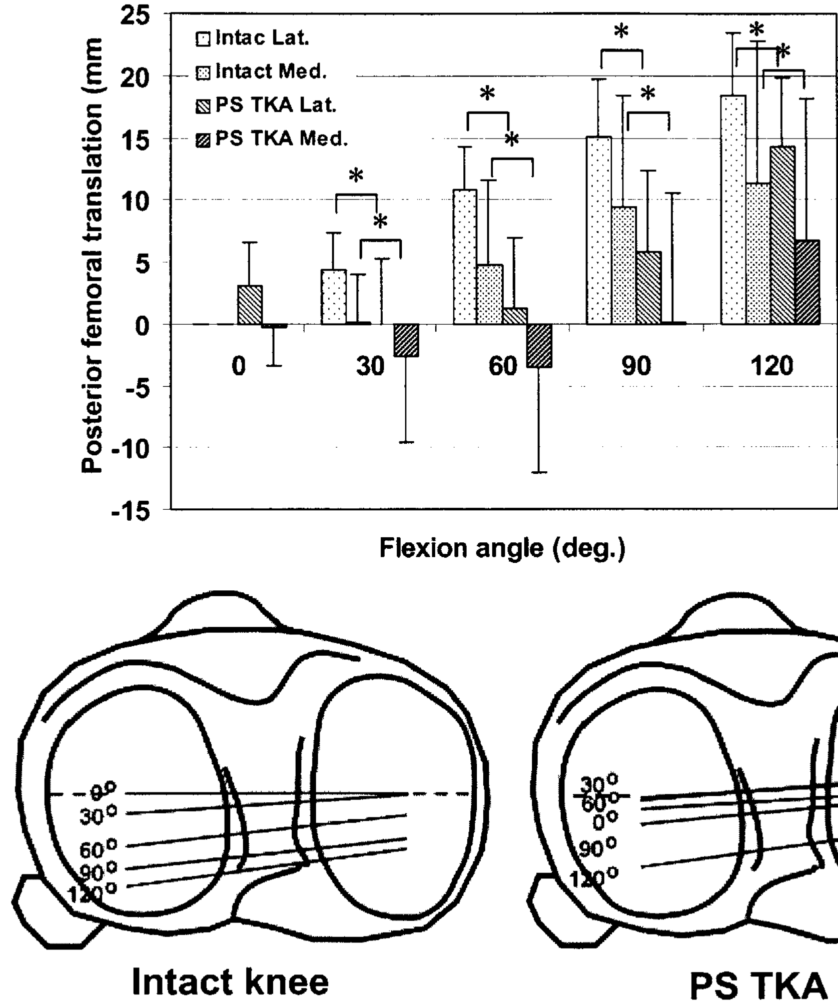


Figure 40: Posterior translation of the lateral and medial femoral condyles on the passive path (* p<0.05). Lat. ≡ lateral; Med. ≡ Medial; PS ≡ posterior-stabilized.

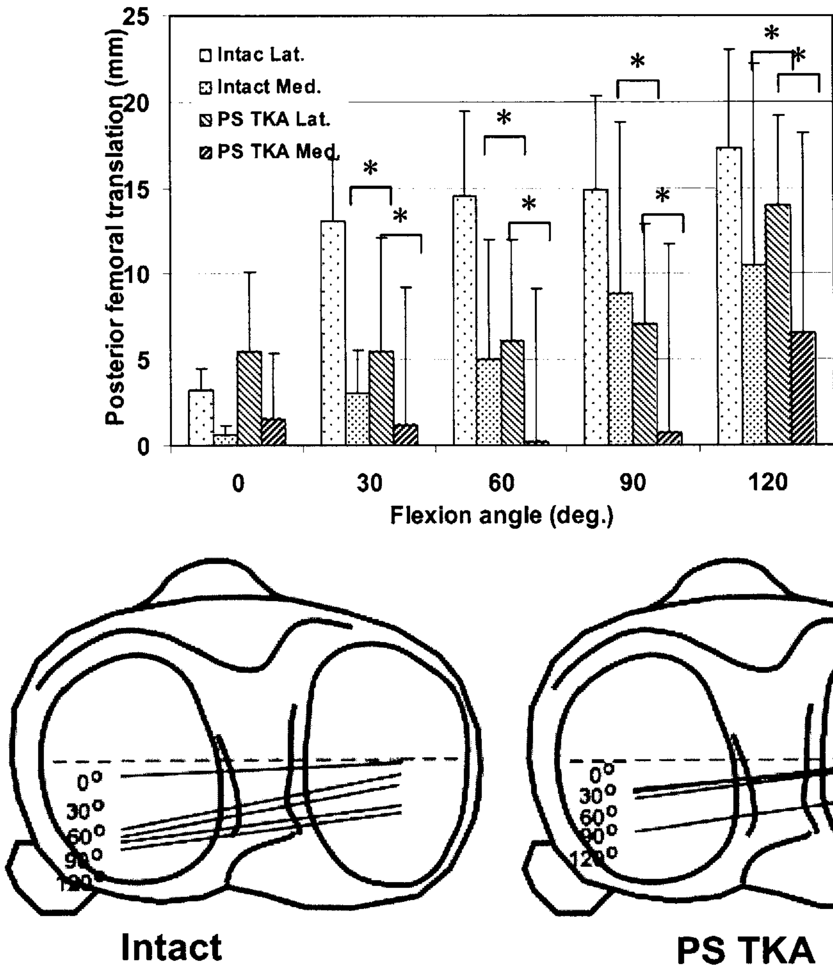


Figure 41: Posterior translation of the lateral and medial femoral condyles under isolated 400N quadriceps load (* $p<0.05$). Lat. ≡ lateral; Med. ≡ Medial; PS ≡ posterior-stabilized.

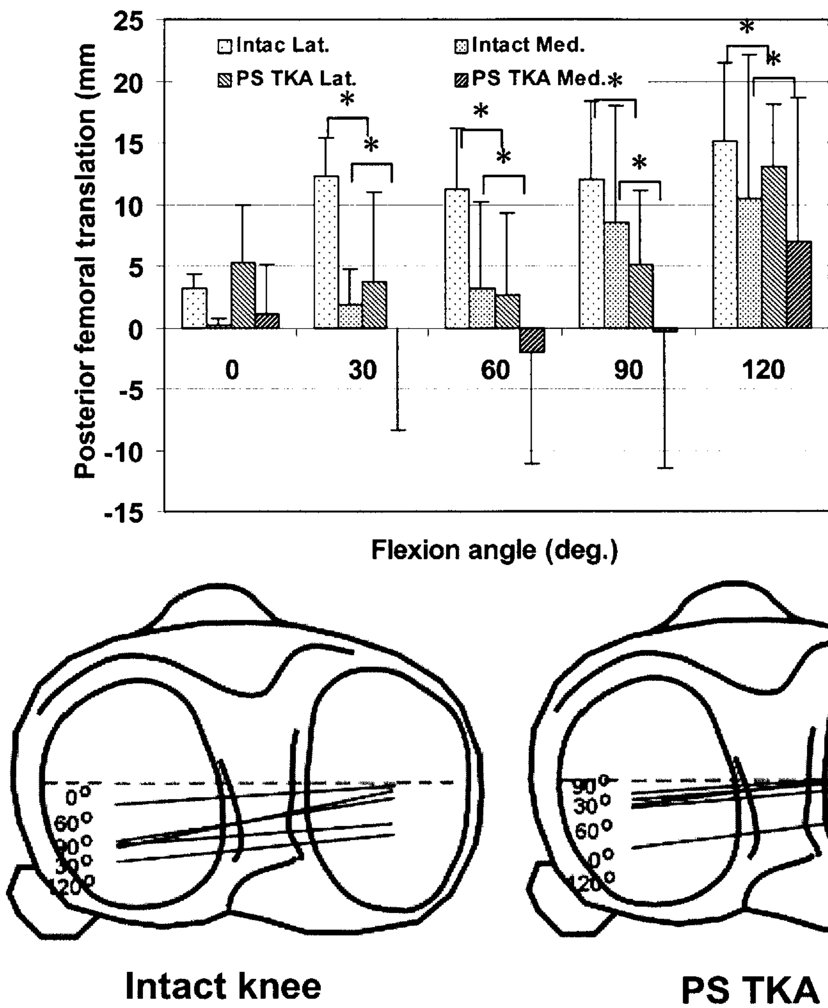


Figure 42: Posterior translation of the lateral and medial femoral condyles under combined 400N quadriceps and 200N hamstrings load (* $p<0.05$). Lat. ≡ lateral; Med. ≡ Medial; PS ≡ posterior-stabilized.

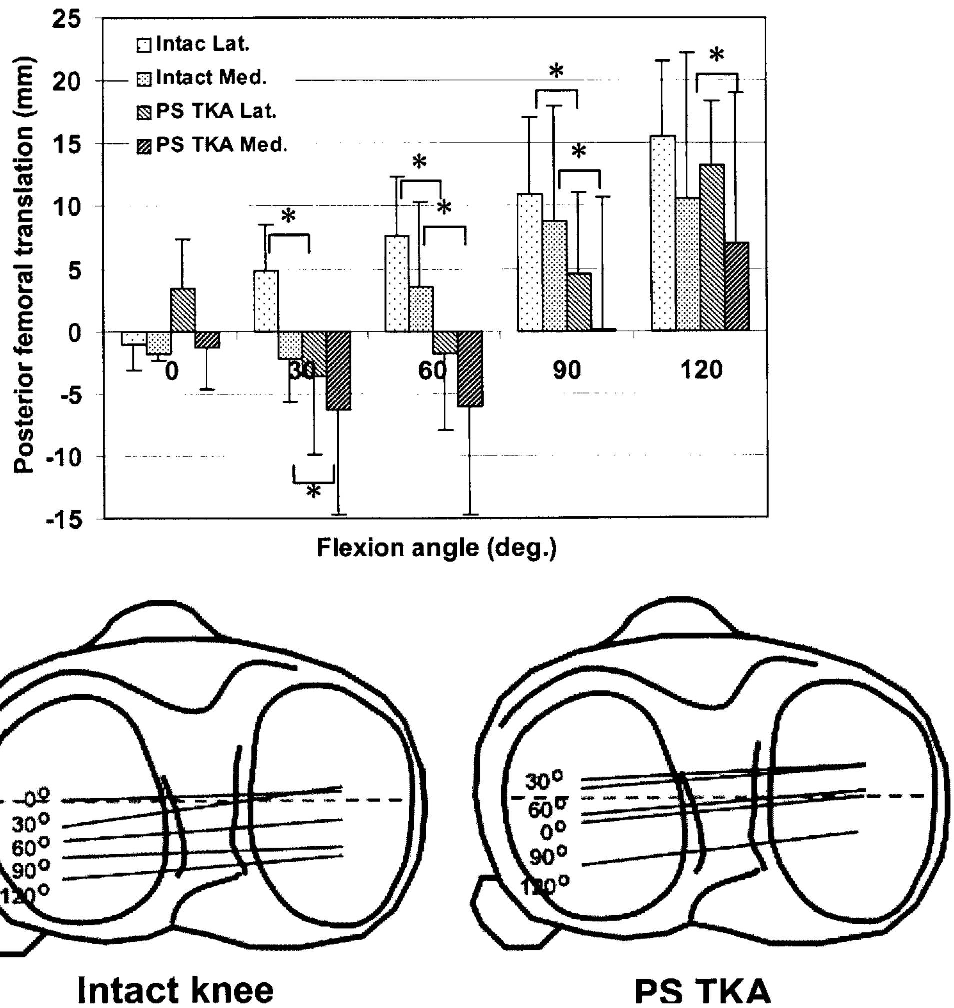


Figure 43: Posterior translation of the lateral and medial femoral condyles under isolate 200N hamstrings load (* p<0.05). Lat. ≡ lateral; Med. ≡ Medial; PS ≡ posterior-stabilized.

6.2.3.2 Cam-Spine Contact Forces

The in-situ contact forces of the femoral cam and tibial spine were approximately 20N between 0° and 60° of flexion under all muscle loading conditions (Figure 44). For the isolated quadriceps load of 400N, the contact force was still about 20N at 90°, but rose to 60N at 120° of flexion. With the addition of 200N hamstrings load, the contact forces increased significantly at 90° and 120° of flexion. The contact force at 90° was

56.7 ± 37.5N and was 82.5 ± 41.7N at 120°. Under the isolated hamstrings load of 200N, the contact forces were 89.3 ± 38.1N at 90° and 82.8 ± 43.0N at 120° of flexion.

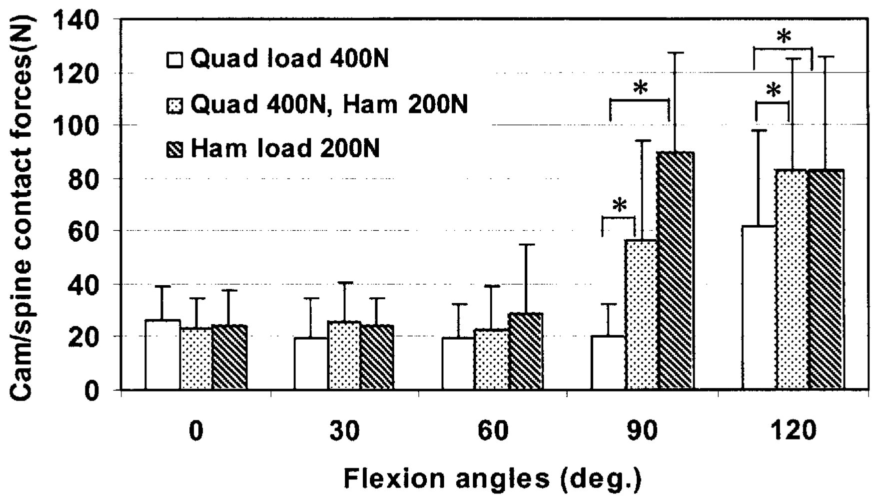


Figure 44: Posterior translation of the lateral and medial femoral condyles under isolate 200N hamstrings load (* $p<0.05$). Lat. ≡ lateral; Med. ≡ Medial; PS ≡ posterior-stabilized.

6.2.4 Conclusions

While the posterior-stabilized TKA was introduced to provide posterior stability and to improve the range of flexion of the knee through the contact mechanism of the cam-spine system at high flexion, no data has been reported on when and how the cam-spine mechanism functions under functional loads. This information is crucial in the development of TKA that will allow knee flexion throughout the entire flexion range. We investigated the effect of the cam-spine system on knee kinematics in a posterior-substituting TKA using an in-vitro experimental set-up. Knee kinematics were measured while the knee was subjected to simulated muscle loads. The native knee kinematics was used as a reference for comparison of TKA kinematics thus minimizing inter-specimen variation.

The results of motion of the native knees at various flexion angles were similar to those reported in the literature [18, 20-22]. All studies showed that maximal posterior femoral translation and internal tibial rotation occurred in the vicinity of 30° of flexion. A reduced posterior femoral translation in posterior-stabilized TKA during step-up and gait was reported in in-vivo studies [10]. Similarly, our data demonstrated that after posterior-stabilized TKA, the knee had a smaller posterior femoral translation in both the lateral and medial condyles under simulated muscle loads compared with the intact knee, except at full extension.

The forces measured at low flexion angles were approximately 20N as seen in Figure 44. Radiographic images taken throughout the entire experiment revealed that the cam-spine mechanism was not engaged in the anterior-posterior direction at low flexion angles, and therefore, the low forces observed. Several factors may contribute to these residual forces. The major factor, we believe, may be the contact of the sides of the spine with the femoral component. The tibial spine was observed to contact the inner wall of the femoral condyles due to tibial rotation and medial-lateral translation before cam-spine engagement in the anterior-posterior direction.

The posterior translations of both the lateral and medial femoral condyles in the posterior-substituting TKA were significantly lower than those observed in the intact knee at all flexion angles except at full extension (Figures 40-43). This posterior-stabilized TKA was designed to have cam-spine contact at approximately 75° of flexion. The cam-spine forces were measured to increase at 90° and 120° under the muscle loads. The data measured in this study was consistent with the component design. Once engagement of the cam and spine took place, the posterior femoral translation was partially restored.

The reduced posterior translation of the femoral condyles after TKA has a profound effect on the capability of the knee to function in high flexion angles. Lower posterior femoral translation may cause early impingement of the femoral shaft with the posterior edge of the tibial component, thus limiting the knee flexion to higher flexion angles. This result may provide an explanation of most current clinical observations of patients' inability to flex higher than 120° [6]. Since the femoral translation is "independent" of muscle loads after cam-spine engagement, it is thus important to manipulate the timing of the cam-spine engagement through surgical options and prosthetic design to enhance the range of knee flexion.

The position of the polyethylene spine on the tibial plateau is an important factor influencing the timing of the cam-spine engagement, as discussed by Delp et al [14] using a two dimensional computer model. The range of posterior translation of the tibia at high flexion of the knee could be manipulated by varying the location of the tibial spine on the tibial plateau. A more anteriorly positioned tibial component will lead to the cam-spine disengaging at higher flexion angles, thus resulting in a lower posterior femoral translation under the muscle loads. A posteriorly positioned component will cause early cam-spine engagement, thus increasing the extent of posterior femoral translation. The discrepancies among the in-vivo measurements cited in the introduction (Section 6.1) might be due to the variation in tibial spine positions in different studies. Different surgical techniques as well as geometrical designs of the components could cause different timing of cam-spine engagement.

In native knees, the differences between posterior translations of the lateral and medial condyles at 30 and 60° of flexion are larger than the differences shown by the posterior-stabilized TKA. This observation indicates that the native knee has a larger internal tibial rotation than the TKA at these flexion angles under the muscle loads. The fact that the TKA resulted in lower internal tibial rotation demonstrated that the TKA was over constrained in rotation at these flexion angles. The reduced internal tibial rotation of the posterior-stabilized TKA may be due to the symmetric geometry of the component. This constraint may adversely affect component fixation. Since most daily activities such as walking involve knee function at low flexion angles, further improvement of TKA should address these constraints.

In summary, in this study, the mechanism of the cam-spine system using a single design posterior-substituting TKA was investigated. It was found that the cam-spine system of this TKA was effective at 90° and 120° of flexion after cam-spine engagement. The engagement of the cam-spine was correlated with an increase in posterior femoral translation in the mid flexion range. The cam-spine contact was shown to not fully facilitate the tibiofemoral rollback at high flexion, a factor that may limit the ability of the TKA from reaching high knee flexion.

6.3 Study 2: High Flexion Fixed and Mobile Posterior-Stabilized TKA (0° - 150° of Flexion)

6.3.1 Introduction

The primary concern of any TKA is to provide the best possible outcome for the patient [1, 23, 24]. The success of any TKA is influenced by a complex interaction between the geometry of the components and the soft tissue envelope that surrounds this articulation. One criterion for judging the success of TKA is the postoperative range of motion achieved. Posterior-stabilizing TKA was introduced to improve knee flexion by promoting posterior femoral rollback. The contact of the femoral cam with the tibial spine prevents excessive posterior tibial translation thus, increasing posterior femoral translation and the range of flexion of the reconstructed knee. Two main designs are currently available: 1. a fixed bearing TKA, where the polyethylene is fixed to the tibial plate and 2. a mobile bearing TKA, whereby the polyethylene is free to move with respect to the tibial plate. However, controversy exists as to the potential benefits of bearing mobility on improving knee flexion.

The concept of mobile-bearing knee implants was introduced in the late 1970's by Goodfellow and O'Connor [25]. They proposed that the mobile-bearing design had better kinematics, range of motion, function, and durability than the existing fixed-bearing knee implants due to its rotating platform. Theoretically, the additional degree of freedom provided by this design should promote load sharing through displacements between the tibial and femoral components, allowing the torque and shear forces of gait to be transferred to the soft tissue in a similar fashion to that which occurs in the native knee.

The ability of the mobile-bearing TKA to restore normal knee kinematics is unclear. Furthermore, it is uncertain as to the exact advantage, if any, that this additional rotation provides, when compared with the fixed-bearing TKA. Therefore, the objective of this study was to investigate posterior femoral translation and tibial rotation in a single design posterior-stabilized TKA offering fixed- and mobile-bearing tibial components (Zimmer Inc, Warsaw, IN) using an in vitro robotic system. The specific research question this study poses is: does the mobile-bearing TKA recreate normal knee translation and rotation better than the fixed-bearing TKA?

6.3.2 Material and Methods

6.3.2.1 Experimental Set-Up

Eleven human knee specimens retrieved post mortem from nine men and two women with an average age of 70 ± 3 years were tested using a robotic testing system described previously [16, 26, 27] (Figure 10 Section 4.3.1). Before experimentation, specimens were thawed overnight at room temperature. Radiographs of each specimen were taken to ensure healthy bone quality. Specimens were hydrated regularly with 0.9% saline solution throughout the experiment to avoid tissue dehydration.

Translation of the lateral and medial condyles and tibial rotation were quantified in this study. We defined the knee center as the center of the transepicondylar line. The centers of the lateral and medial femoral condyles were defined as a point located 25 mm proximal to the joint line and located on either side of the knee center (Figure 11 Section

4.3.1) [26, 27]. The zero position was defined as the location of the condyle when the intact knee was at full extension.

6.3.2.2 Testing Protocol (Figure 45)

The intact knee was tested first. After specimen installation, the robot determined the passive path from 0° to 150° flexion using a force-moment control algorithm [26]. Three variations of muscle loads were simulated in this study: (1) isolated quadriceps force of 400 N; (2) combined quadriceps and hamstrings load (400 N and 200 N, respectively); and (3) isolated hamstring force of 200 N. The kinematics of the intact knee under these simulated muscle loads were measured at selected flexion angles (0°, 30°, 60°, 90°, 120°, and 150°).

An orthopaedic surgeon then performed knee replacement surgery, implanting a fixed-bearing posterior-stabilized TKA on the knee (LPS-Flex, Zimmer Inc., Warsaw, IN). The surgical procedure was described in detail in Chapter 5. The anterior and posterior cruciate ligaments were resected. The patella was not resurfaced. Trial components were inserted and knee stability, passive range of motion, patella tracking, and flexion and extension gaps were evaluated. The arthrotomy was closed with continuous suture, as was the skin.

After insertion of the TKA components, a new passive path was determined. The kinematics of the fixed-bearing posterior-stabilized TKA under the same muscle loads then was recorded at 0°, 30°, 60°, 90°, 120°, and 150° of flexion.

Components from the first TKA then were removed and prior bony cuts were modified as indicated to allow insertion of a mobile-bearing posterior-stabilized TKA (LPS-Mobile, Zimmer Inc., Warsaw, IN). This procedure was done on the same knee and by the same surgeon. The femoral component remained unchanged. The tibial component was replaced with a mobile-bearing TKA base plate. Subsequently, a mobile-bearing polyethylene liner was inserted. The arthrotomy and the skin were closed with continuous suture. No detectable degradation of the bone or soft tissue was observed. A new passive path was defined for the mobile-bearing posterior-stabilized TKA followed by experiments to determine kinematics under predefined conditions of muscle loading at the same selected flexion angles.

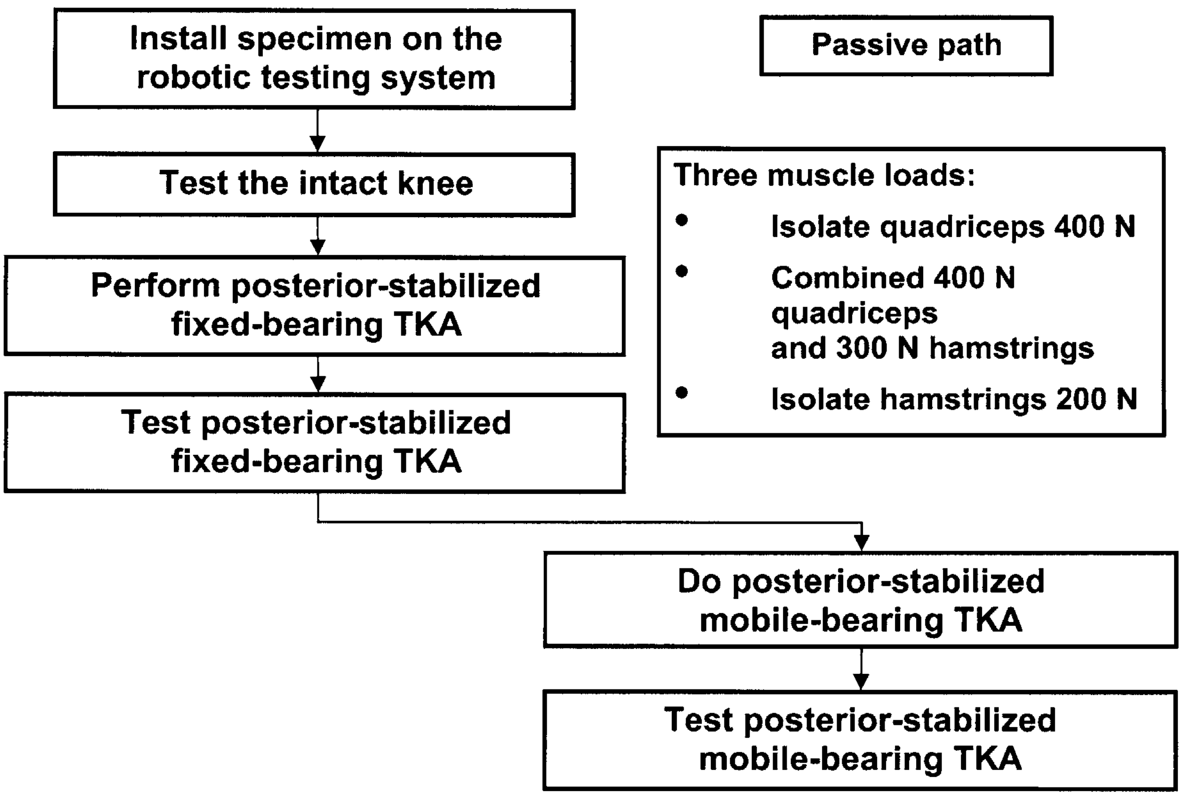


Figure 45: Overall testing protocol for high flexion fixed and mobile posterior-stabilized TKA experiment.

6.3.2.3 Component Design

The LPS-Flex and the LPS-Mobile TKAs share the same femoral component (Figure 46). The component features symmetric anatomic radii (medial and lateral condyles) in the sagittal plane with a larger distal radius and a smaller posterior radius, and includes a cam-spine mechanism. The component has a deep patellar groove that has been extended more distally and proximally than the traditional posterior-stabilized TKA component. This allows the patella to be supported fully at as much as 85° flexion.

The tibial articular surface of the LPS-Flex system is dished in the frontal and sagittal plane to match the corresponding geometries of the femoral component. The fixed-bearing knee system is designed to accommodate ±12° rotation between the femoral and tibial articular surfaces.

The tibial articular surface of the LPS-Mobile system includes a smaller frontal radius of curvature than the sagittal radius of curvature. The mobile-bearing polyethylene is more congruent than the fixed-bearing TKA. The pivot axis of the articular surface is located anteriorly. The tibial plate includes a rotational stop that helps to prevent spinout of the articular surface. The mobile-bearing knee system is designed to accommodate ±25° rotation between the tibial plate and the polyethylene. In this study, the LPS-Mobile bearing knee allows solely for pure rotation of the tibial plateau. Neither AP translation nor unconstrained motions, as seen in other designs, were allowed [28].

6.3.2.4 Statistical Analysis

A two-way repeated measures analysis of variance (ANOVA) was used to analyze the effect of knee structure modification on joint motion. Joint motion included lateral and medial posterior femoral translation and tibial rotation. A Newman-Keuls test was done to detect statistical significance between knee states (intact knee, fixed-bearing posterior-stabilized TKA, and mobile-bearing posterior-stabilized TKA). Significance level was set at $p < 0.05$.

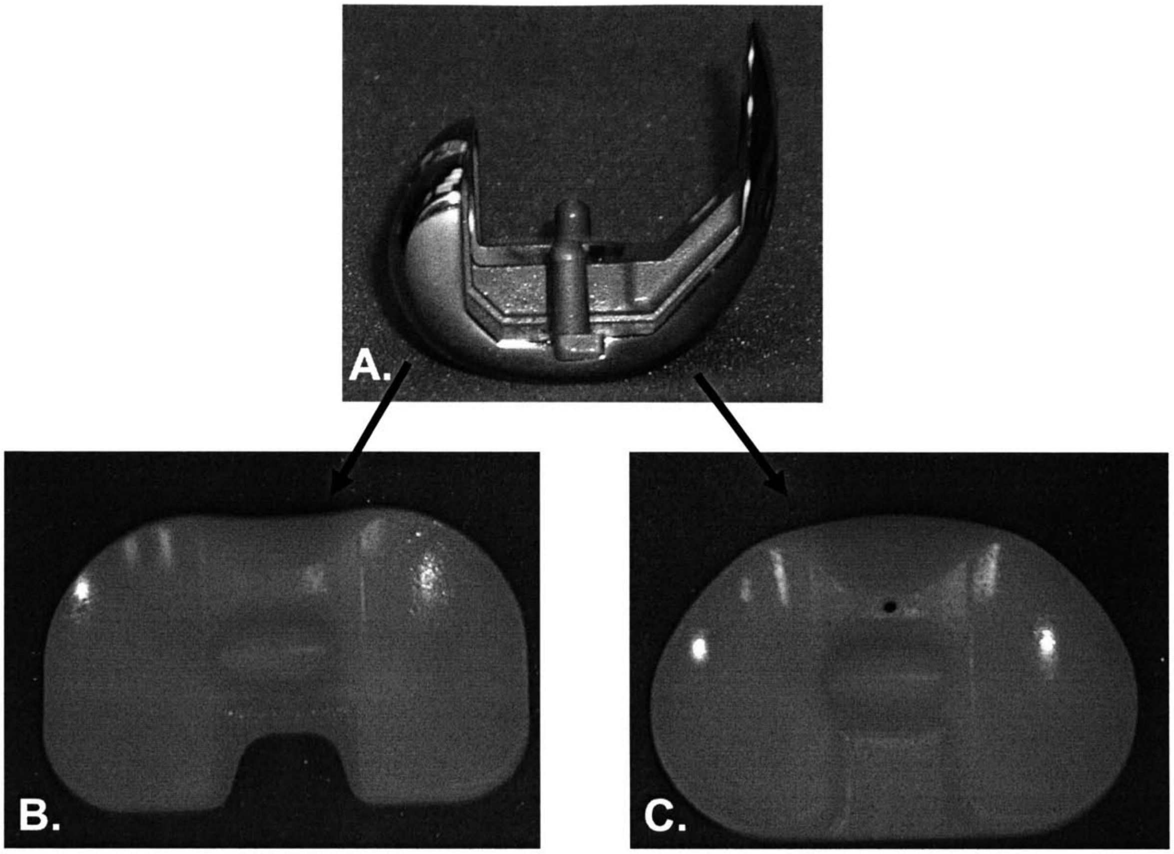


Figure 46: Photograph shows the A) femoral component, B) fixed-bearing polyethylene liner, and C) mobile-bearing polyethylene liner used in this study.

6.3.3 Results

6.3.3.1 Femoral Translation during Passive Knee Motion

Minimal rollback was observed for the intact knee below 60° flexion. The lateral femoral condyle center of the intact knee (Figure 47A) translated posteriorly 1.9 ± 2.7 mm at 30° flexion. Posterior translation of the medial femoral condyle (Figure 47B) was minimal below 30° flexion (0.3 ± 2.1 mm). Between 30° and 150° flexion, posterior femoral translation increased gradually for the medial and the lateral condyles reaching a maximum of 24.9 ± 9.1 mm and 34.6 ± 10.4 mm at 150°, respectively. Minimal posterior femoral translation was observed for the fixed- and the mobile-bearing TKAs at flexion angles of less than 90°. At higher degrees of flexion, posterior translation of the lateral condyle of both TKA variants increased, reaching a maximum of approximately 30 mm at 150° flexion (Figure 47A). Posterior femoral translation of the medial condyle of both TKAs was less than that recorded for the intact knee for all flexion angles. Approximately 85% of the intact knee rollback was recovered on the medial side at 150° of flexion (Figure 47B).

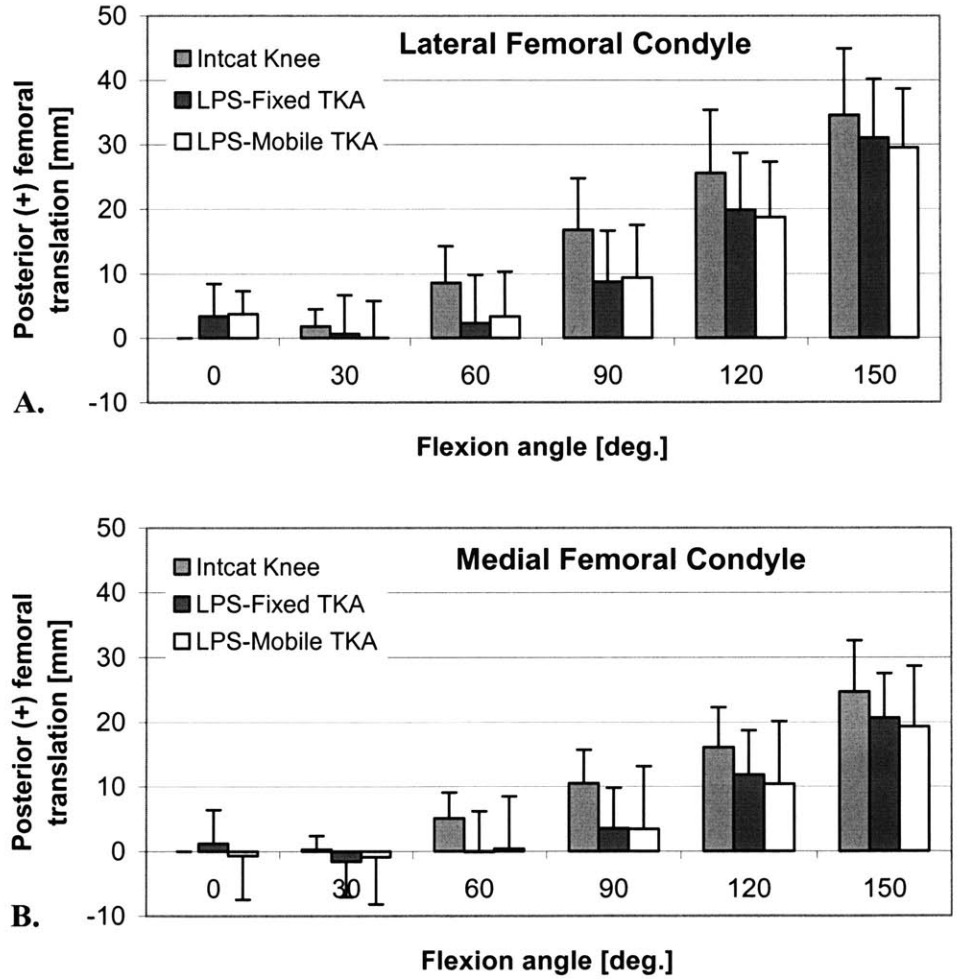
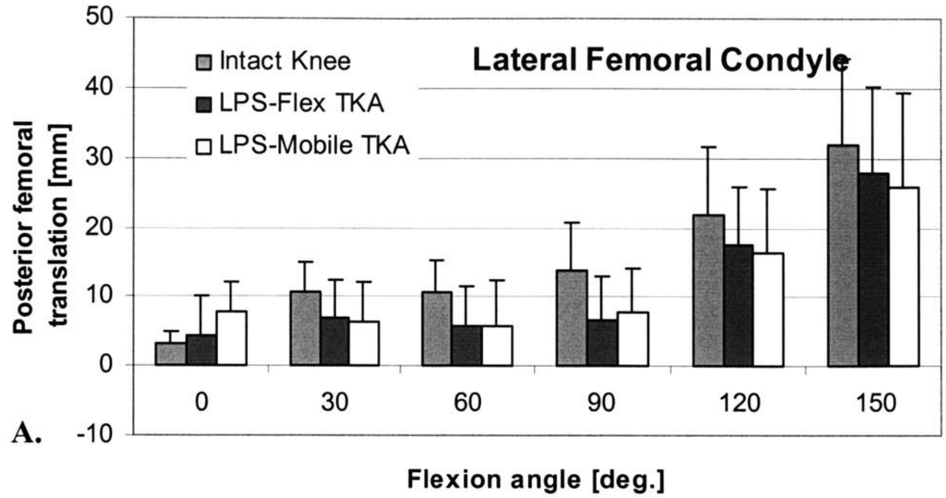


Figure 47: Graphs show the posterior translation of the A) lateral and B) medial femoral condyles on the passive path.

6.3.3.2 Femoral Translation under Combined Muscle Loads

Under combined muscle load, the lateral femoral condyle of the intact knee translated posteriorly 10.8 ± 4.3 mm at 30° of flexion and this increased gradually to 31.9 ± 12.5 mm at 150° (Figure 48A). Fixed- and mobile-bearing total TKAs restored 60% of the femoral rollback observed in the native knee at 30°. The fixed-bearing lateral condyle translated 27.9 ± 12.2 mm at 150°. A similar result was observed for the mobile-bearing lateral condyle (25.9 ± 13.3 mm) at 150°. On the medial side, the femoral condyles of the intact knee translated 0.7 ± 3.5 mm at 30° and 22.9 ± 11.3 mm at 150°. The femoral condyles of the fixed-bearing arthroplasty translated 1.7 ± 7.3 mm at 30° and 20.4 ± 9.9 mm at 150° (Figure 48B). The mobile bearing arthroplasty exhibited translations which were similar to the fixed-bearing knee replacement (2.1 ± 7.4 mm at 30° and 19.3 ± 13.5 mm at 150°).



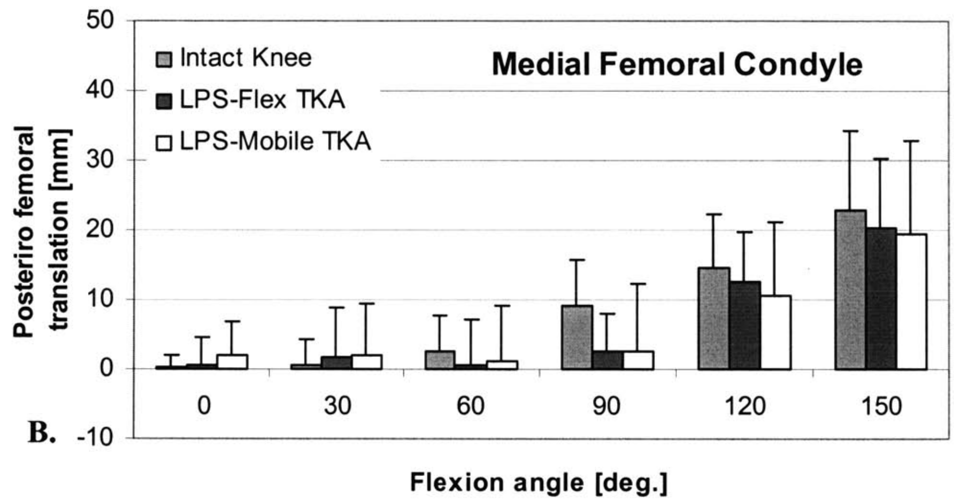


Figure 48: Graphs show the posterior (+) translation of the A) lateral and B) medial femoral condyles under combined muscle load.

6.3.3.3 Tibial Rotation during Passive Knee Motion

For all knees, internal tibial rotation increased with increasing flexion angle (Figure 49). In the native knee, at 30° of flexion, the tibia internally rotated 1.8° ± 2.3°, reaching a maximum of 11.5° ± 7.9° at 150° of flexion. After reconstruction with the LPS-Fixed-bearing TKA, the tibia rotated internally 2.6° ± 6.8° at 30° of flexion and the amount of rotation remained relatively unchanged for flexion angles less then 90°.

Thereafter, internal tibial rotation increased gradually with increasing flexion angle reaching a maximum value of 12.2° ± 8.9° at 150°. There was no significant difference ($p>0.05$) in terms of the amount of internal tibial rotation between the intact knee and the LPS-Fixed bearing TKA throughout the range of passive knee motion.

The LPS-Mobile-bearing TKA showed a similar trend in terms of internal tibial rotation, similar to that of the LPS-Fixed-bearing TKA. After LPS-Mobile TKA reconstruction, the tibia externally rotated 1.1° ± 6.8° at 30° flexion. Beyond 30°, the tibia increasingly rotated internally with increasing flexion angle. Maximum internal tibial rotation was observed at 150° (12.0° ± 9.9°). There was no significant difference ($p > 0.05$) in terms of internal tibial rotation between the intact knee and the LPS-Mobile TKA throughout the range of motion along the passive path. Furthermore, there was no significant difference ($p > 0.05$) in terms of tibial rotation between the LPS-Fixed TKA and LPS-Mobile throughout the entire range of motion along the passive knee motion.

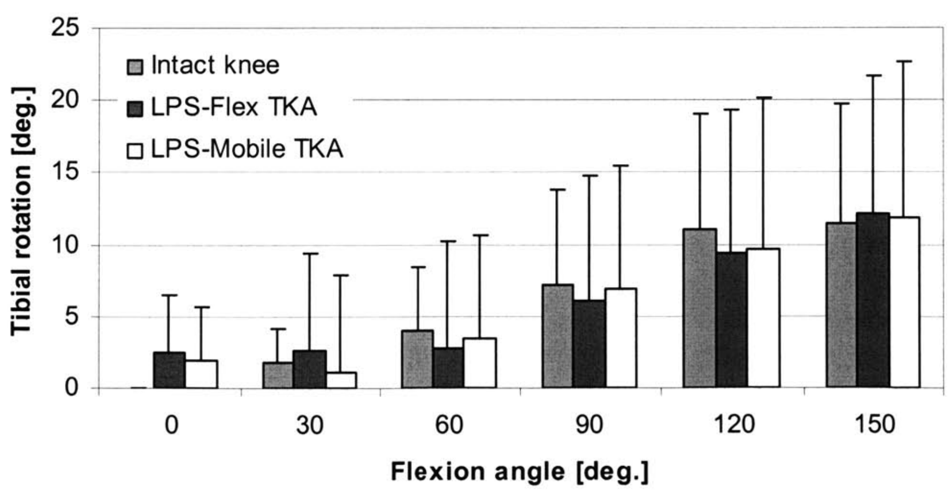


Figure 49: Graph shows the internal tibial rotation as a function of knee flexion on the passive path.

6.3.3.4 Tibial Rotation under Combined Muscle Loads

For all knees, internal tibial rotation under combined muscle loads increased with increasing flexion angle (Figure 50). The intact knee internally rotated 11.0° ± 5.7° at 30° and 10.4° ± 7.3° at 150° of flexion. Fixed-bearing and mobile-bearing TKAs restored approximately 45% of intact tibial rotation at 30° of flexion, respectively. At 150°, the fixed-bearing TKA rotated internally 8.8° ± 8.3°, restoring 85% of intact knee rotation, whereas the mobile-bearing TKA rotated internally 7.7° ± 8.4°, restoring 74% of intact knee internal rotation. As with internal tibial rotation on the passive path, there was no statistical significant difference ($p > 0.05$) in terms of internal tibial rotation between the intact knee and the two TKAs. Furthermore, there was no significant difference ($p > 0.05$) in terms of tibial rotation between the LPS-Fixed and LPS-Mobile TKAs throughout the entire range of motion along the passive knee motion.

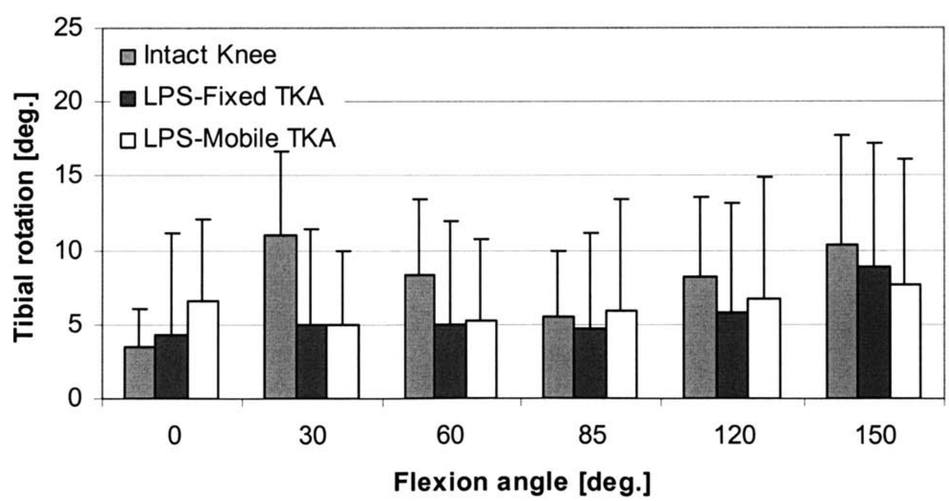


Figure 50: Graph shows the internal tibial rotation as a function of knee flexion under combined muscle load.

6.3.4 Conclusions

This study compares the kinematics of the posterior-stabilized TKA (fixed- and mobile-bearing designs) with those of the intact knee using robotic technology. At present, the kinematics of mobile-bearing knee replacements are poorly understood. However, despite this knowledge shortfall, most orthopaedic total joint manufacturing companies have developed or currently are developing mobile-bearing total knee prostheses.

The data reported here indicates that posterior femoral translation occurs under conditions of muscle loading and in the absence of muscle loading (passive path) in the native knee. These finding are consistent with those of prior experiments [29]. Posterior femoral translation was also observed after reconstruction for fixed- and mobile-bearing TKAs. However, fixed- and mobile-bearing posterior-stabilized TKAs only partially restored the posterior femoral translation observed in the native knee. For example, at 150° of flexion, both design types restored approximately 85% of the posterior translation observed in the native knee. For all knees, the lateral condyle was observed to translate further posteriorly than the medial condyle, suggesting coupled translation and rotation. It was observed that the medial femoral condyle first pivoted at less than 60° of flexion and then rolled back (translation). A closer examination of femoral rollback indicates that the cam-spine interaction in both TKA designs seems to contribute to femoral rollback at flexion angles greater than 75°. This is shown by a consistently observed increase in posterior femoral translation above this flexion angle.

In the native knee, an increase in flexion angle corresponded with an increase in internal tibial rotation. Similarly, after TKA, internal tibial rotation was observed with increasing flexion, but only partial restoration of this motion occurred. The fixed-bearing posterior-stabilized TKA restored 85% of the internal tibial rotation observed in the native knee at 150° of flexion, whereas the mobile-bearing posterior-stabilized TKA recovered 74% of the native knee's internal tibial rotation at the same flexion angle.

Quantification of the amount of rotation that actually occurs between the femur and the tibial insert versus the tibal insert and the tibial tray is of great interest. We have shown that the mobile tibial insert rapidly reaches its stop at < 90° flexion (Figure 51). After this point, the TKA is essentially transformed into the equivalent of a fixed-bearing implant. This conclusion is supported firmly by our data, as we observed no statistically significant kinematic differences between the fixed- and the mobile-bearing TKA.

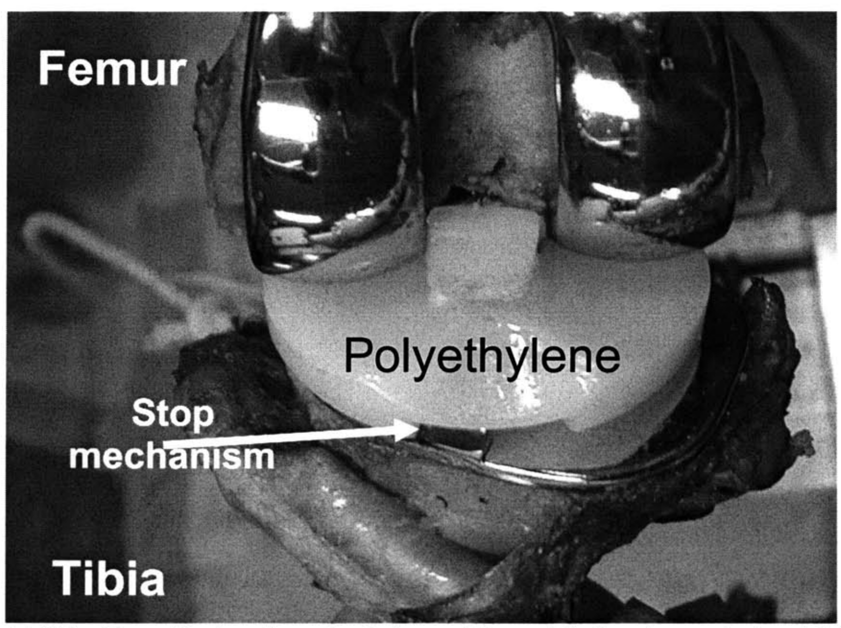


Figure 51: At 90° of knee flexion, the polyethylene of the mobile bearing posterior-stabilized TKA reached the anterior stop.

We observed that once the polyethylene liner in the mobile bearing posterior-stabilized TKA reaches the stop, it also overhangs, exceeding the tibial plate margin (Figure 52). We reported that with increasing flexion, the posterior condyles translate posteriorly contacting with the proximal aspect of the posterior edge of the polyethylene. However, as the distal aspect of the posterior edge of the polyethylene liner is unsupported at high knee flexion, fracture and polyethylene failure is most likely to occur.

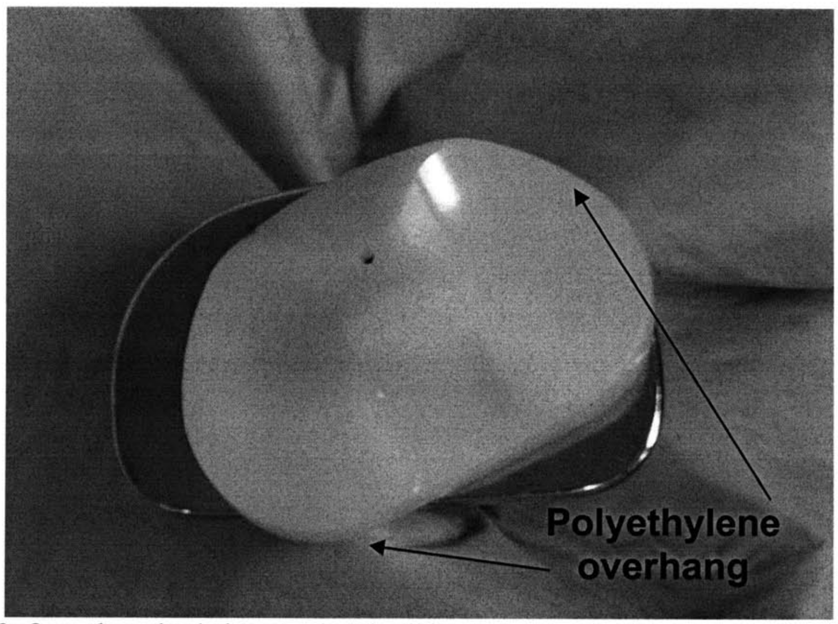


Figure 52: Once the polyethylene reaches the tibial stop, polyethylene overhangs the tibial tray.

Extensive research has been done on fixed-bearing TKA [9, 26, 29-33]. Fixed-bearing TKAs provide durable long-term fixation with prosthetic survival rates of 95% to 97% reported at 10 to 15 years [6, 34, 35]. Good clinical results and comparable range of motion of as much as 120° has been reported for the fixed-bearing TKA [9, 30-33]. However, some fixed-bearing designs have had problems with polyethylene wear jeopardizing long-term survivorship [34-36].

In a multi center analysis, Dennis et al [33] determined the in vivo femorotibial contact patterns for various TKAs using video fluoroscopy and a fitting technique. They reported that from full extension to 90° flexion, posterior femoral translation occurred at the lateral condyle, whereas the medial femur was approximately stationary. They observed that the posterior-cruciate-substituting TKA routinely showed less posterior femoral rollback than that observed in normal knees. Data from our study are similar to the fluoroscopic data.

Limited information regarding the mobile-bearing TKA is available [24, 28, 34, 37-42]. For example, even though one implant system, the LCS rotating-platform design (DePuy Orthopedics, Warsaw, IN), has been in clinical use for the past 20 years, few studies have reported on the long-term results of mobile-bearing TKA [28, 37, 40, 41].

Existing analyses regarding the kinematics of rotating platform TKAs have shown numerous abnormalities relative to results from studies of the native knee [6, 41, 43]. Fluoroscopic analysis of PCL-sacrificing rotating platform TKA has revealed paradoxical anterior femoral translation with flexion in the range of 60° to 90° [41]. Rotational abnormalities and femoral liftoff primarily of the lateral condyle also have been shown as has a reduction in overall knee flexion relative to patients with fixed-bearing posterior-stabilized TKA [6, 43].

D'Lima et al [38] compared the kinematics of fixed- and mobile-bearing cruciate-retaining and cruciate-substituting TKAs (Press-Fit-Condylar, DePuy Johnson & Johnson, Inc, Warsaw, IN) using the Oxford rig. Four different variations of the TKA design were implanted sequentially on the same cadaver and tested between 0° and 90° flexion. They reported that all TKAs significantly altered the kinematics of the native knee. All of the designs exhibited rollback. They showed that the rotating-bearing TKA had minimal effect on tibiofemoral kinematics but may provide an increase in contact area and a reduction contact stress. Their results are in agreement with the results reported in this study.

Mobile-bearing TKAs initially were developed to decrease the contact stresses on polyethylene and to reduce polyethylene wear. These prostheses were designed to create a dual-surface articulation such that the polyethylene insert that articulates with the metallic femoral component would also have a source of kinematic freedom at the upper surface of the metallic tibial tray on which it rests. The purpose of this feature was to reduce surface and subsurface stress states at the bearing surfaces and at the bone-implant surfaces specifically by maximizing the conformity of the tibial and femoral components and allowing mobility of the bearing surface [28]. The theory behind maximizing articular surface conformity is that polyethylene contact stress is inversely proportionate to the contact area for a given load such that more conforming tibial components increase contact area and therefore reduce contact stresses [1, 16, 40-42, 44-47]. In the fixed-bearing knee design, increased conformity can result in transfer of excessive stresses to the implant-bone interface, which has been associated with loosening of the tibial component [48, 49].

6.4 Discussion

Ideally, the goal of modern TKA designs should be to restore native knee function as closely as possible. The amount of posterior femoral translation (femoral rollback) is directly proportional to and highly essential for knee flexion. This obligatory motion creates a space for the tibia posterior to the femur, thereby preventing posterior impingement and allowing knee flexion.

The robotic testing system provides us with the ability to sequentially test a native knee followed by various posterior-stabilized TKA under the same muscle loads on the robot eliminates inter-specimen variability. Knee kinematics specifically addressed by the two studies included femoral translation and tibial rotation of the knee before and after reconstruction. Furthermore, the robotic set-up allows for assessment and measurements of the function and forces of the cam-spine mechanism.

Our study shows that normal knee kinematics are altered after the implantation of the various posterior-stabilized TKA used in both studies. All designs partially restore normal knee kinematics on the passive path of knee motion and under simulated muscle loads. In all knees, greater translation of the lateral condyle was observed when compared with the medial condyle, signifying coupled translation and rotation. Although a direct comparison between the two studies is difficult, we concluded that all prosthetic types had similar kinematic patterns regarding posterior femoral translation and tibiofemoral rotation despite the fact that their designs are different.

The cam-spine interaction appears to contribute to femoral rollback in posterior-stabilized TKA. Posterior femoral translation and cam-spine contact seems to occur at a similar flexion angle, suggesting that the cam-spine system supplements for the function of the posterior cruciate ligament. At low flexion, the femoral cam and the tibial spine were not engaged (Figure 53). Minimal cam-spine forces at flexion angles less than 75° correlated with little rollback. At approximately 75°, engagement of the cam-spine occurred at the same time where increase in posterior femoral rollback was observed.

The engagement continued until 135° of flexion. At that point, disengagement (Figures 54 and 55) was noticed. Interestingly, despite the separation of the cam-spine, posterior femoral translation continued to rise in the high flexion designs.

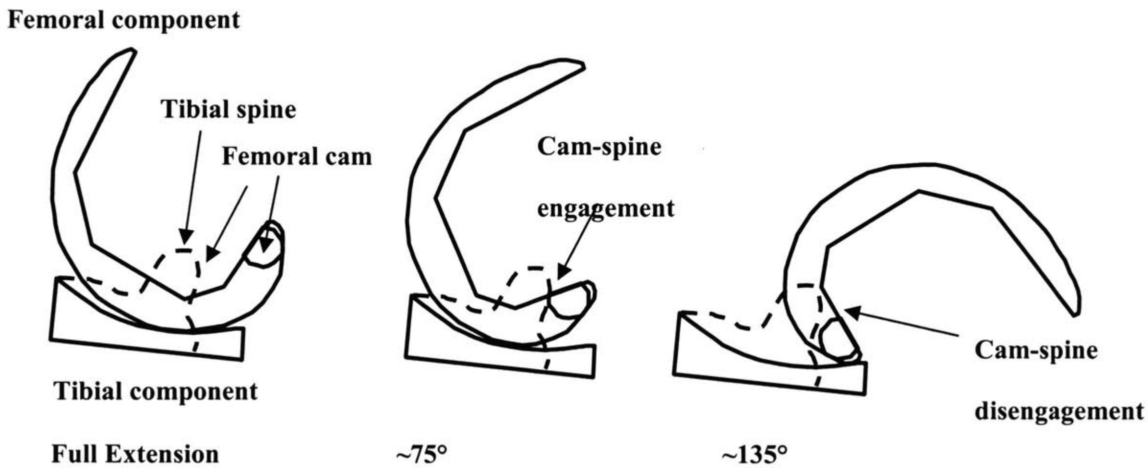


Figure 53: The figure shows the three main stages of the cam-spine.

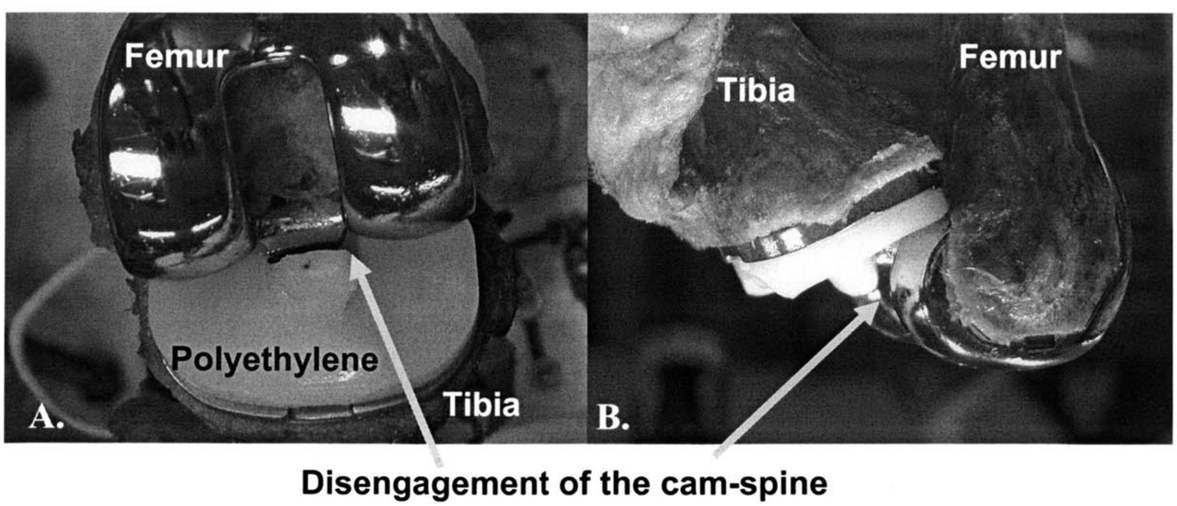


Figure 54: The disengagement of the cam-spine in the fixed bearing posterior-stabilized TKA was observed at high knee flexion: A) anterior view and B) medial view.

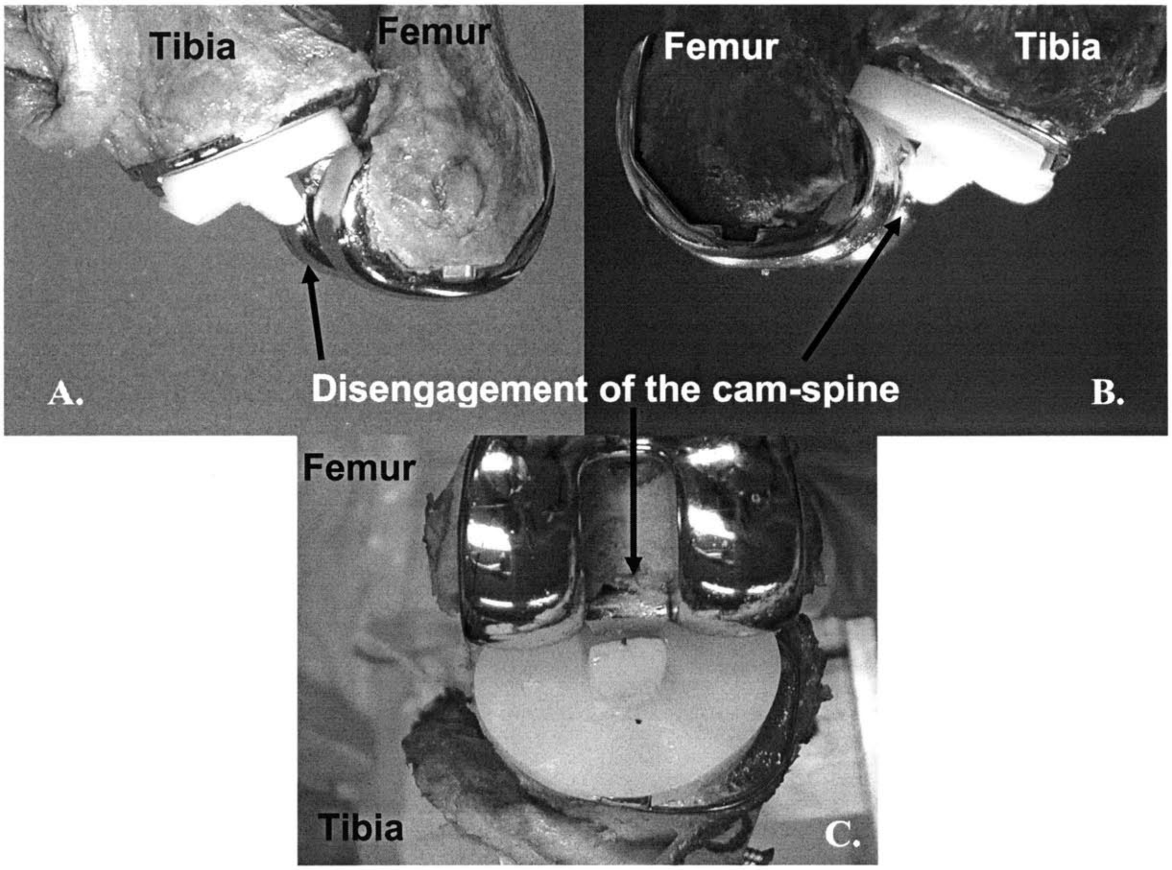


Figure 55: The disengagement of the cam-spine in the mobile bearing posterior-stabilized TKA was observed at high knee flexion: A) medial view and B) lateral view, and C) anterior view.

The comparable kinematics observed in all TKA designs suggests that soft tissue around the knee may play a critical role in constraining knee motion particularly for the high flexion design. The menisci in the native knee are integral components of knee motion especially at higher flexion angles. On the medial side, the meniscus surrounds the joint enhancing articular congruity. On the lateral side, at lower degrees of flexion (< 110°), the meniscus behaves similarly to the medial meniscus. However, as flexion angles increase beyond 110°, the lateral meniscus slides off of the tibia posteriorly, creating a mobile surround support system for the joint, allowing the lateral femur to roll off the back of the tibia. Because the menisci are removed during arthroplasty reconstruction, this unique function of the menisci is not recovered by either TKA design.

However, we observed that the soft tissue compression (hamstrings muscle, fat, skin, posterior capsule) play a crucial role in guiding knee motion at high knee flexion.

During the different experiments, we observed the effect of tibial tray placing on the tibiofemoral contact at high knee flexion. Currently, orthopaedic surgeons align the tibial tray with the anterior aspect of the tibia as shown in Figures 56 and 57. Sizing of the tibial component is done based on the amount of bone it covers. In many cases, smaller component completely cover one side of the tibia while the other side is only partially covered. For example, as seen in Figure 56, the lateral tibial plateau is fully covered by the tibial tray while several mm of bone remain exposed on the medial side. Larger component will not be beneficial as it will completely cover the larger side (the medial tibial plateau in Figure 57) but will extend beyond the smaller side (lateral tibial plateau) impinging with the femoral shaft at high knee flexion. One possible solution is to place the tibial tray a bit more posteriorly, or even aligning the posterior edges of the tibial tray with the tibia to provide larger contact area and potentially improved knee stability at high flexion angles (Figure 57).

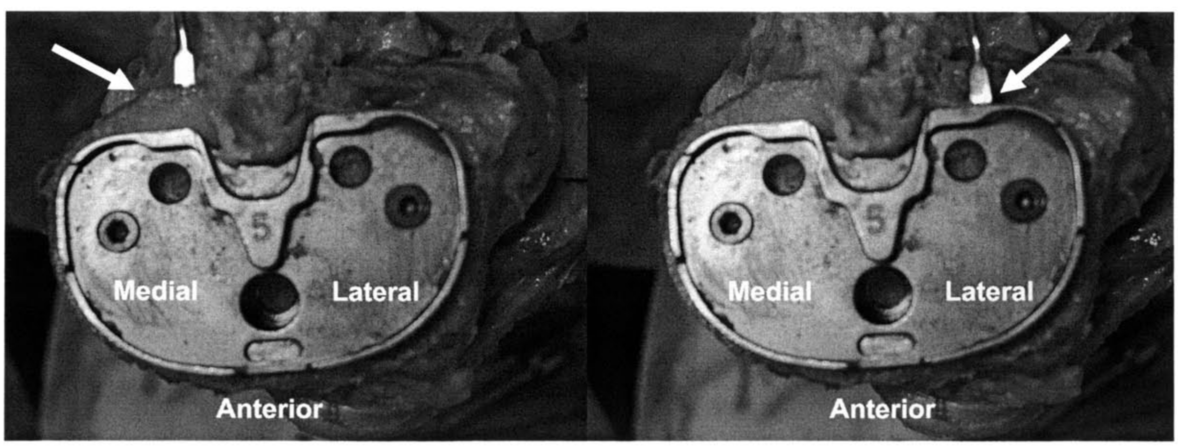


Figure 56: The sizing and placement of the tibial tray. On the lateral side, the posterior edges of the tibial tray and the tibial bone are aligned. However, on the medial side, the tibial tray does not fully cover the medial face of the tibial bone.

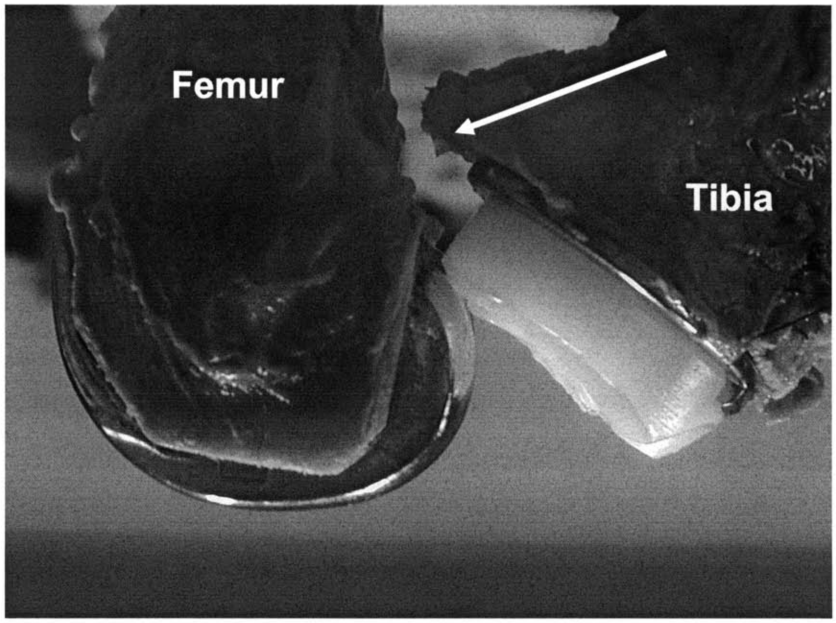


Figure 57: Moving the tibial tray posteriorly may provide greater contact area and improved stability at high knee flexion.

The quantitative data presented in our study may serve as an aid in the development of a rationale for additional improvement in surgical techniques and prosthesis design, so that normal knee function over the entire range of flexion may be restored.

6.5 References

1. Stern, S.H. and J.N. Insall, *Posterior stabilized prosthesis. Results after follow-up of nine to twelve years.* J Bone Joint Surg Am, 1992. 74(7): p. 980-6.
2. Insall, J.N., P.F. Lachiewicz, and A.H. Burstein, *The posterior stabilized condylar prosthesis: a modification of the total condylar design. Two to four-year clinical experience.* J Bone Joint Surg Am, 1982. 64(9): p. 1317-23.
3. Ranawat, C.S., et al., *Long-term results of the total condylar knee arthroplasty. A 15-year survivorship study.* Clin Orthop, 1993(286): p. 94-102.
4. Emmerson, K.P., C.G. Moran, and I.M. Pinder, *Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up.* J Bone Joint Surg Br, 1996. 78(3): p. 441-5.
5. Aglietti, P. and R. Buzzi, *Posteriorly stabilised total-condylar knee replacement. Three to eight years' follow-up of 85 knees.* J Bone Joint Surg Br, 1988. 70(2): p. 211-6.
6. Dennis, D.A., et al., *Range of motion after total knee arthroplasty: the effect of implant design and weight-bearing conditions.* J Arthroplasty, 1998. 13(7): p. 748-52.
7. Stiehl, J.B., et al., *Fluoroscopic analysis of kinematics after posterior-cruciate-retaining knee arthroplasty.* J Bone Joint Surg Br, 1995. 77(6): p. 884-9.
8. Banks, S.A., G.D. Markovich, and W.A. Hodge, *In vivo kinematics of cruciate-retaining and -substituting knee arthroplasties.* J Arthroplasty, 1997. 12(3): p. 297-304.
9. Dennis, D.A., et al., *In vivo knee kinematics derived using an inverse perspective technique.* Clin Orthop, 1996(331): p. 107-17.
10. Tarnowski, L.E., et al., *Three-dimensional motion of cruciate retaining and posterior stabilized TKAs during walking.* Trans ORS, 1998. 44: p. 804.
11. Andriacchi, T.P., et al., *A point cluster method for in vivo motion analysis: applied to a study of knee kinematics.* J Biomech Eng, 1998. 120(6): p. 743-9.

12. Essinger, J.R., et al., *A mathematical model for the evaluation of the behaviour during flexion of condylar-type knee prostheses.* J Biomech, 1989. 22(11-12): p. 1229-41.

13. Wimmer, M.A. and T.P. Andriacchi, *Tractive forces during rolling motion of the knee: implications for wear in total knee replacement.* J Biomech, 1997. 30(2): p. 131-7.

14. Delp, S.L., J.H. Kocmond, and S.H. Stern, *Tradeoffs between motion and stability in posterior substituting knee arthroplasty design.* J Biomech, 1995. 28(10): p. 1155-66.

15. Piazza, S.J., et al., *Posterior tilting of the tibial component decreases femoral rollback in posterior-substituting knee replacement: a computer simulation study.* J Orthop Res, 1998. 16(2): p. 264-70.

16. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

17. Rudy, T.W., et al., *A combined robotic/universal force sensor approach to determine in situ forces of knee ligaments.* J Biomech, 1996. 29(10): p. 1357-60.

18. Li, G., et al., *The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL.* J Biomech, 1999. 32(4): p. 395-400.

19. Li, G., et al., *Prediction of antagonistic muscle forces using inverse dynamic optimization during flexion/extension of the knee.* J Biomech Eng, 1999. 121(3): p. 316-22.

20. Hirokawa, S., et al., *Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction.* Am J Sports Med, 1992. 20(3): p. 299-306.

21. Hoher, J., et al., *In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study.* J Orthop Res, 1999. 17(5): p. 763-8.

22. Torzilli, P.A., X. Deng, and R.F. Warren, *The effect of joint-compressive load and quadriceps muscle force on knee motion in the intact and anterior cruciate ligament-sectioned knee.* Am J Sports Med, 1994. 22(1): p. 105-12.

23. Anouchi, Y., et al., *Range of motion in total knee replacement.* Clin Orthop, 1996. 331: p. 87-92.
24. Callaghan, J., et al., *Mobile-Bearing knee replacement: concepts and results.* J Bone Joint Surg Am, 2000. 82(7): p. 1020.
25. Goodfellow, J. and J. O'Connor, *The mechanics of the knee and prosthesis design.* J Bone Joint Surg Br, 1978. 60-B(3): p. 358-69.
26. Li, G., et al., *Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads--an in vitro experimental study.* J Orthop Res, 2002. 20(4): p. 887-92.
27. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.
28. Callaghan, J.J., et al., *Cemented rotating-platform total knee replacement. A nine to twelve- year follow-up study.* J Bone Joint Surg Am, 2000. 82(5): p. 705-11.
29. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.
30. Aglietti, P., R. Buzzi, and A. Gaudenzi, *Patellofemoral functional results and complications with the posterior stabilized total condylar knee prosthesis.* J Arthroplasty, 1988. 3(1): p. 17-25.
31. Aglietti, P., et al., *An analysis of the survival rate of total-condylar total knee prostheses with posterior stability.* Ital J Orthop Traumatol, 1988. 14(4): p. 419-28.
32. Dennis, D., et al., *Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation.* Clin Orthop, 1992. 281: p. 168-176.
33. Dennis, D.A., et al., *In vivo anteroposterior femorotibial translation of total knee arthroplasty: a multicenter analysis.* Clin Orthop, 1998(356): p. 47-57.
34. Colizza, W.A., J.N. Insall, and G.R. Scuderi, *The posterior stabilized total knee prosthesis. Assessment of polyethylene damage and osteolysis after a ten-year-minimum follow-up.* J Bone Joint Surg Am, 1995. 77(11): p. 1713-20.

35. Insall, J.N., et al., *The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements.* J Bone Joint Surg Am, 1983. 65(5): p. 619-28.

36. Wasielewski, R.C., et al., *Tibial insert undersurface as a contributing source of polyethylene wear debris.* Clin Orthop, 1997(345): p. 53-9.

37. Buechel, F.F., Sr., et al., *Twenty-year evaluation of meniscal bearing and rotating platform knee replacements.* Clin Orthop, 2001. 388: p. 41-50.

38. D'Lima, D.D., et al., *Comparison between the kinematics of fixed and rotating bearing knee prostheses.* Clin Orthop, 2000(380): p. 151-7.

39. Kaper, B.P., et al., *Medium-term results of a mobile bearing total knee replacement.* Clin Orthop, 1999(367): p. 201-9.

40. Stiehl, J.B., *World Experience with low Contact Stress mobile bearing Total Knee Arthroplasty: A literature Review.* Orthopedics, 2002. February, Supplement.

41. Stiehl, J.B., et al., *In vivo kinematic comparison of posterior cruciate ligament retention or sacrifice with a mobile bearing total knee arthroplasty.* Am J Knee Surg, 2000. 13(1): p. 13-8.

42. Vertullo, C.J., et al., *Mobile Bearing In Primary Knee Arthroplasty.* Journal of American Academy of orthopedic Surgeons, 2001. 9(6): p. 355-64.

43. Dennis, D., et al., *In Vivo femoral condylar lift-off in total knee arthroplasty.* Orthop Trans, 1997. 21: p. 1112.

44. Bartel, D.L., et al., *The effect of conformity and plastic thickness on contact stresses in metal-backed plastic implants.* J. Biomech. Eng., 1985. 107: p. 193-199.

45. Matsuda, S., et al., *Contact stress analysis in meniscal bearing total knee arthroplasty.* J Arthroplasty, 1998. 13(6): p. 699-706.

46. Miller, M.C., et al., *The effect of component placement on knee kinetics after arthroplasty with an unconstrained prosthesis.* J Orthop Res, 2001. 19(4): p. 614-20.

47. Stiehl, J.B., et al., *In vivo kinematic analysis of a mobile bearing total knee prosthesis.* Clin Orthop, 1997(345): p. 60-6.

48. Bryan, R.S. and J.A. Rand, *Revision total knee arthroplasty.* Clin Orthop, 1982(170): p. 116-22.

49. Werner, F., D. Foster, and D. Murray, *The influence of design on the transmission of torque across knee prostheses.* JBJS, 1978. 60A: p. 342-348.

Chapter 7

THE ANALYSIS OF THE FLEXION GAP BEFORE AND AFTER POSTERIOR-STABILIZED TKA

7.1 Introduction

Total knee arthroplasty (TKA) has evolved into one of the most effective and reliable surgical procedures performed today. The evolution of TKA has entailed progress in patient selection, surgical technique, component design and post-operative rehabilitation. Many independent reports of good or excellent long-term outcomes with a wide variety of contemporary TKA designs are published [1-9].

It is known that the human knee can flex up to 160° of flexion [10, 11]. Knee flexion is essential for daily living, and the amount of flexion has been linked to functional outcome [12]. For example, to squat or kneel, an individual requires as much as 160° of flexion [13], and bathtub use requires approximately 130° of knee flexion [14].

Although many reports document flexion after contemporary TKA of 100° to 115°, deep flexion after arthroplasty surgery has not been consistently attained [15]. Furthermore, the vast majority of research conducted on knee kinematics before and after TKA has focused on flexion angles less than 120°. Hence, the biomechanical principles of knee flexion both in the intact and TKA have been well described for flexion angles less than 120°, but our understanding of these principles in deep/high flexion is limited to relatively few studies [11, 16, 17].

One concept with potential importance in achieving deep flexion is flexion/extension gap balancing. In this technique, the surgeon aims to equalize the extension space at 0° with the flexion space at 90° [18-20]. This concept is founded on the hypothesis that, in the intact knee, these joint spaces are equal throughout the range of motion between 0° and 90°. Therefore, total knee kinematics and function will be most successful if it is able to optimally restore this normal kinematics behavior. However, no studies to date quantify the flexion or extension gap in deep flexion either in the intact or the reconstructed knees. This study uses robotic technology and an *in vitro* experimental design in order to achieve two aims: First, we attempt to provide the first quantitative analysis of the flexion gap in deep flexion for the intact knee. Second, we attempt to quantify and perform a comparative analysis of the flexion gap for a TKA using new components designed to achieve high flexion (NexGen LPS-Flex, Zimmer Inc., Warsaw, IN).

7.2 Materials and Methods

7.2.1 Specimen Preparation

Seven fresh-frozen human cadaver knee specimens (average age 67 ± 11 years; range 56 to 85 years old; 4 female knee 4 male knees, 4 right knees and 4 left knees) were thawed overnight at room temperature. Each knee included approximately 25cm of femoral shaft and 25cm of tibial shaft, and all the soft tissue around the knee joint (hamstrings muscles, skin, posterior capsule, ligaments, fat, and menisci). Prior to experimentation, radiographic images of each specimen were taken to exclude specimens with previous injuries or surgeries. Each specimen was manually flexed from full extension to 150° of flexion to assure full range of motion. The specimens were hydrated regularly with 0.9% saline solution to avoid tissue dehydration.

7.2.2 Kinematics Analysis

Each specimen was manually pre-conditioned ten times prior to its installation on the robotic system [21-24]. The intact knee was tested first. The passive knee motion from full extension (0°) to 150° of flexion was determined by the robotic system [21-24].

An orthopaedic surgeon then performed a posterior-stabilized TKA (NexGen® LPS-Flex, Zimmer Inc., Warsaw, IN) on the same knee. The "surgery was performed through a midline skin incision followed by a medial parapatellar arthrotomy. The femoral cuts were performed with the guidance of an intramedullary alignment system to determine varus/valgus alignment. The epicondylar axis was used as a guide for rotational alignment. The femoral component was placed in 5° of valgus and in 3° of external rotation. The tibial cut was performed using an extramedullary system, with the tibial crest and the center of the tibial plateau serving as reference points. The tibia was cut with a 7° posterior slope, and the tibial component was ultimately aligned with the junction of the medial and middle thirds of the tibial tuberosity. The posterior cruciate ligament was resected. The patella was not resurfaced. Trial components were inserted and knee stability, passive range of motion, patella tracking, and flexion and extension gaps were evaluated. The arthrotomy was closed with continuous suture, as was the skin" [25]. Following the insertion of the femoral, tibial and polyethylene components, a new passive path was determined for the reconstructed knee.

7.2.3 Knee Models

The skin and all soft tissue were than dissected away from the specimen. Each specimen was visually inspected to assure good soft tissue quality and no cartilage degeneration. Three non-collinear points were selected on the femoral shaft and three on the tibial shaft as reference positions for the bones in space (Figure 5 in Chapter 3). The pre-recorded passive path for both knees was digitally recorded through the use of a Microscribe® 3DX (Immersion Corporation, San Jose, CA) at every five degrees from full extension to 150° of flexion using the reference points. The tibial and femoral surface contours were then digitized. The data was imported into Rhinoceros® (Robert McNeel & Associates, Seattle, WA), a 3D rendering program.

For each knee, the femoral and tibial bones were scanned using 3D x-ray technology (Iso^{3D} C-arm, Siemens, Germany). The sagittal images (Figure 58A) of the bones were imported into MATLAB®. A MATLAB® code for outlining the contours of the bones was written (Figure 58B). The contours where then imported into Rhinoceros® and three-dimensional bone models were created (Figure 58C). The completed 3D models were then matched to the previously recorded kinematics using the bone contours and the reference points.

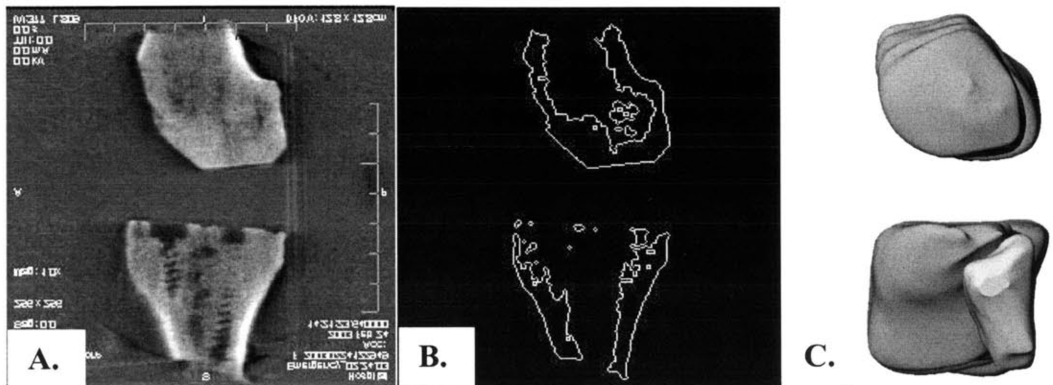


Figure 58: Knee model: A) sagittal image from x-ray, B) image after MATLAB® edge detection, and C) the final 3D model.

7.2.4 Flexion Gap Measurement

Identifiable points were selected on both the femur (Figure 59A) and the tibia (Figure 59B) to assure consistency in the distance measurements. The flexion and extension gaps were measured by measuring the average perpendicular distances from these reproducible points at the bone-components interface on both the medial and lateral femoral condyles and the tibial plateau (Figure 59C). The gap was measured at every 5° under both the intact and the TKA passive knee kinematics.

A two-way repeated measures Analysis of Variance (ANOVA) was used to detect whether knee state and flexion angle had a significant effect of the dimensions of the gap. Newman-Keuls post hoc tests were performed to detect statistical significance between knee states and flexion angles. Significant level was set to $p < 0.05$.

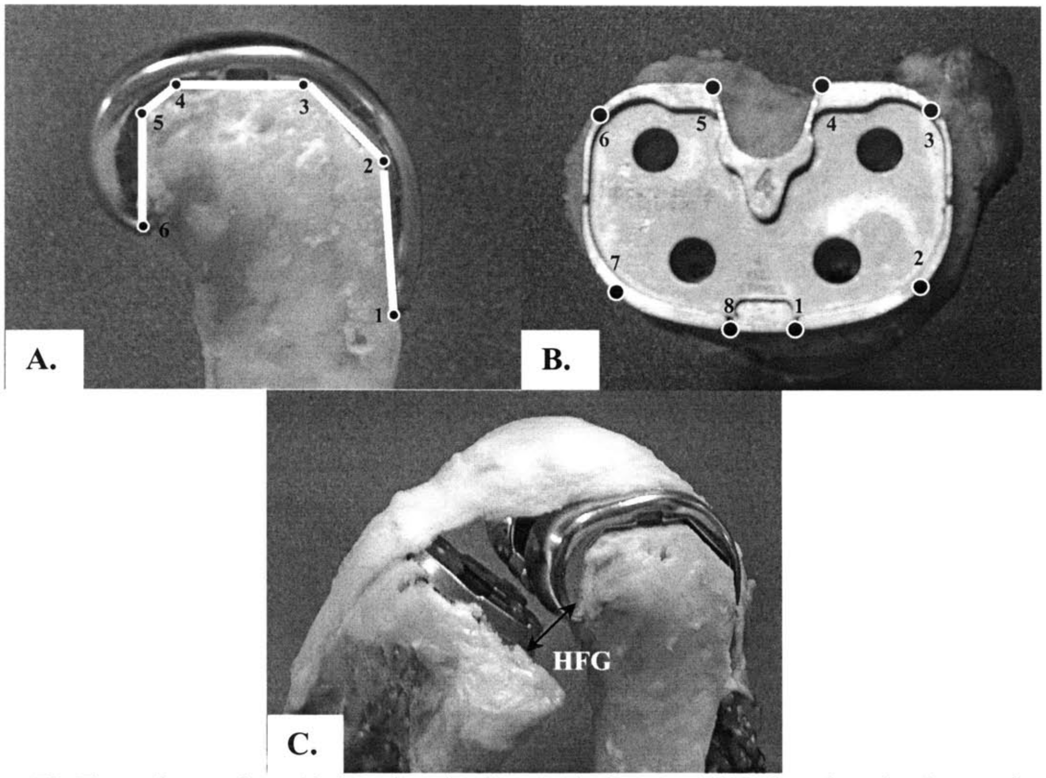


Figure 59: These figures show A) the selected points on the femur, and B) the selected points on the tibia, used in the gap analysis. C) The high flexion gap (HFG) was defined as the shortest distance between the femoral and the tibial bones

7.3 Results

7.3.1 Flexion/Extension Balancing of Intact and TKA at Full Extension (0°) and 90° of Flexion

All of the knees in this study were balanced intra-operative at 0° and 90° degrees. No soft tissue release was performed in any specimen. The native knee and the TKA at 0° had extension gaps of 19.2 ± 2.2 mm (mean ± std) and 21.9 ± 2.0 mm, respectively. At 90° of flexion, the flexion gaps of the native and TKA were 23.1 ± 2.3 mm and 25.0 ± 2.5 mm, respectively. No statistical significant differences between the lateral and the medial gaps were observed for both knees and at both flexion angles ($p>0.05$), signifying that all knees were balanced throughout the entire range of motion. The difference between the extension and flexion gaps at 0 and 90° when measured quantitatively for the native knee and the TKA were 2.7 ± 1.7 mm and 1.9 ± 1.9 mm, respectively.

7.3.2 Flexion Gap of Intact and TKA in Deep Knee Flexion

Beyond 90° of flexion, the gap for both the intact and the TKA knees progressively closed as the flexion angles approached deep flexion of 140° (Figure 60). However, with further increase in knee flexion beyond 140°, a change in concavity occurred, and the gap dimension increased.

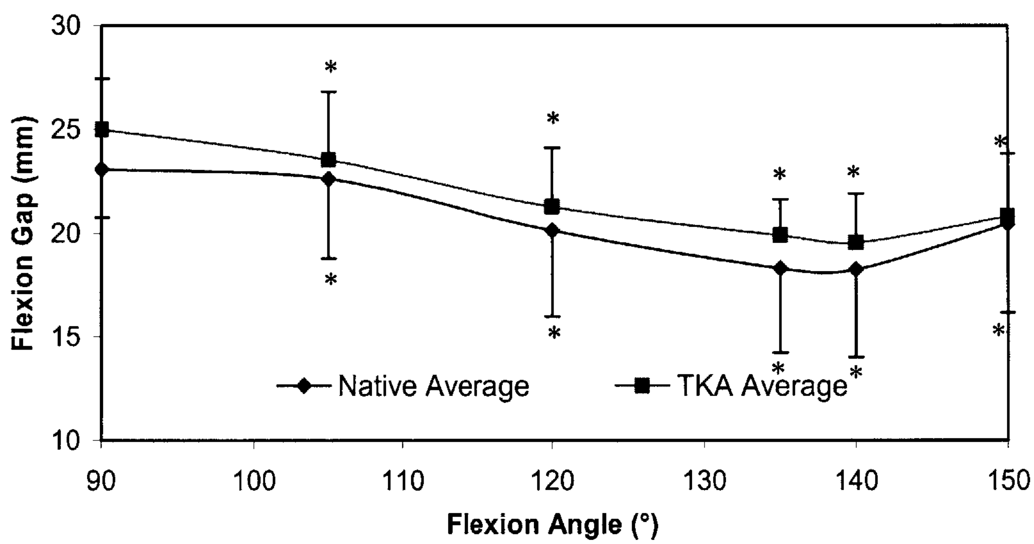
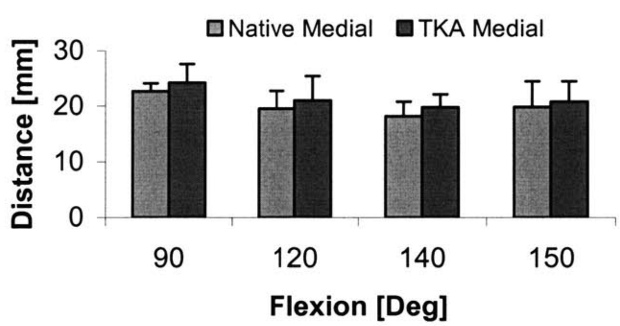


Figure 60: The flexion gap for both knees progressively decreased with increasing knee flexion (* denotes $p < 0.05$ when the given flexion angle is compared to the same knee state at 90°).

At 120° of flexion, the average distance (gap) between the femur and the surface of the tibial plateau was 20.1 ± 4.2 mm in the native knee. At the same flexion angle, the average flexion gap for the reconstructed knee was 21.3 ± 2.8 mm. At 140° of flexion, the intact and reconstructed knees' gap closed even further reaching a value of 18.2 ± 4.2 mm and 19.5 ± 2.4 mm, respectively. With further increase in knee flexion to 150°, the intact knee gap opened (increased) to 20.4 ± 4.3 mm. Similar behavior was noted for the reconstructed knee where its gap widens at 150° to 20.8 ± 3.0 mm. No statistical significant differences between the intact and the reconstructed knee beyond 90° was noted ($p>0.05$). However, there was a statistical significant differences between the values of the knees at 90° and the values at any other flexion angle greater than 90° ($p<0.05$).

To assure that the knee was balanced at high flexion angles, the medial and lateral gaps were measured. Figure 61 outlines the narrowing of the medial and lateral flexion gaps for both knee states, at selected flexion angles, as a function of knee flexion. The decrease in flexion gap with increasing flexion between 90° and 140° followed by an increase beyond 140° of flexion was observed in both condyles. There was no significant difference between the medial and lateral gap throughout the entire range of flexion.

A.



B.

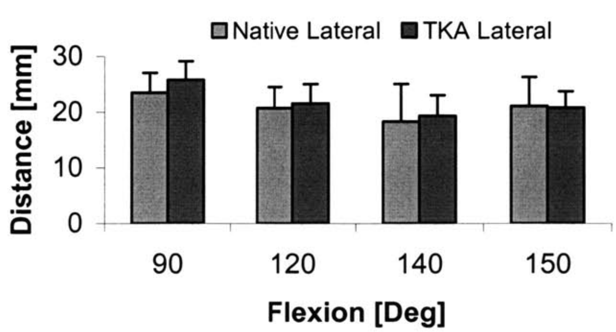


Figure 61: The A) medial and B) lateral flexion gaps for both knees at selected flexion angles.

7.4 Discussion

Although TKA is very effective at alleviating pain and restoring satisfactory levels of function, many patients after contemporary TKA have not been able to restore deep flexion, and, therefore, compromising their normal function [10, 12, 16, 26].

There are many factors that may contribute to limited knee flexion after TKA. Studies have shown that preoperative knee range of motion is the most important factor in influencing postoperative knee flexion although diagnosis, deformity, age, sex, and patient weight are all considered important factors [27-32]. Intra-operatively, soft tissue balancing, tibial slope, patella resurfacing and tracking, posterior cruciate ligament management, wound closure, component position and sizing as well as prosthetic design may all influence knee motion [15, 33, 34]. Post-operatively, rehabilitation including continuous passive motion and specific physical therapy protocols may also play a role in influencing flexion after surgery [15, 35, 36].

However, only few studies in the literature describe the mechanisms that are particularly relevant to deep flexion [11, 16, 37-39]. Belleman et a. [38], used computer-aided design videofluoroscopy to study deep flexion in 150 patients after cruciate-retaining TKA's. They documented impingement of the posterior aspect of the tibial insert against the shaft of the femur in deep flexion and determined that the loss of posterior femoral offset after TKA resulted in decreased maximal flexion postoperatively. In their analysis, for every millimeter lost in posterior condylar offset as compared with the each knee preoperatively, there was a mean loss of 6.1° of terminal flexion. Patients that lost more than 3mm of posterior condylar offset had a mean 29.7 less maximal flexion after TKA as compared with those in whom posterior condylar offset was restored to within 3 mm. Nakagawa et al [39], using MRI, examined the tiobiofemoral movement in 20 male adults in full knee flexion. They reported that at deep flexion, the tibia plateau and the femoral condyles have lost contact and that the contact seems to be on the posterior horn of the menisci.

Limited data is available regarding the factors of currently available TKA components that limit patients from flexion beyond 120° after TKA. Most of our understanding of the biomechanical mechanisms of knee motion results from both *in vivo* and *in vitro* studies investigating motion below 120° of flexion. One factor that is thought to play a role in limiting further flexion is the flexion gap. This study represents the first attempt to define and characterize the dimensions of the flexion gap beyond 90°. Furthermore, this study is the first to quantitatively define and analyze the flexion gap in deep flexion for both the native knee and TKA using contemporary components designed to maximize deep flexion.

Currently, surgeons only balance the knee at 0° and at 90° [18, 20] and consider the knee balanced when the gaps are of equal dimension. The balancing procedure includes a level of surgical intuition and is limited by current technology (spacers). The accuracy of the flexion extension gaps in 104 posterior-stabilized total knee replacements was evaluated by Griffin et al [18] who reported a maximum of 3mm difference between the flexion and extension gaps.

Surgeons consider the intact knee to be the gold standard for ligament/soft tissue balancing. This study revealed no significant differences between the intact and the reconstructed knees throughout the entire range of flexion, suggesting that the TKA behaves as the native knee defines 'balanced'.

In this study, we found that beyond 90°, the flexion gap decreased with increasing flexion for both the native and the reconstructed knees until approximately 140°. With further flexion beyond 140°, the dimension of the flexion gap increased. We attribute the increase in gap size at deep flexion to the compression of the posterior soft tissue. As the knee flexes into deep angles, the posterior soft tissue, including the skin, hamstrings muscles, posterior capsule, and intact menisci, are compressed and distorted between the femoral and tibial shafts until no further compaction is possible. At that point, the posterior soft tissue pushes the tibial shaft anteriorly opening the flexion gap. As described in chapter 4, this soft tissue compression promotes posterior femoral translation at high knee flexion, which allows patients to achieve higher knee flexion.

It has been suggested that surgeons considered increasing the tibial slope as a solution for a knee that is tight in flexion. It is believed that increasing the tibial slop allows patients to achieve high knee flexion. However, increasing the tibial slop willl alo open (increase) the flexion gap after TKA. This study shows that narrowing of the flexion gap at high knee flexion occurs in both the intact and the reconstructed knees. In fact, there was no statistical significant difference between the intact and the reconstructed gap throughout the entire range of flexion. The gap associated with this TKA design appeared to play a minimal role in limiting the knee from achieving high flexion.

This study only examined one type of high flexion TKA design (LPS-Flex, Zimmer, Warsaw, IN). We showed that component design had minimal effect of the ability of the knee to achieve high flexion angles. Other factors potentially affecting knee flexion should be investigated (e.g. soft tissue constraint, tibiofemoral contact, oversized femoral component). Furthermore, it s known that partial or complete release of the posterior cruciate ligament and soft tissue increases the extension gap in the intact knee [40]. An investigation on the effect of these releases on TKA and the high flexion gap should be performed to better understand the behavior of knees at high knee flexion. Future direction in the analysis of the flexion gap should also include the effect of muscle loads on the gap dimensions.

This is a fist attempt to define and quantify the flexion gap at high knee flexion. Currently, surgeons only balance the knee at 0° and 90°. Good clinical results are reported for TKA in this flexion range. However, one major problem that has not been addressed fully is that patients do not gain high degrees of flexion after TKA, regardless the fact that patients express interest in achieving higher knee flexions after the surgery. The data reported here suggest that the high flexion gap concept is only one factor that possibly plays a role in limiting high degrees of knee flexion. For the design and bony landmarks used in this study, we concluded that the flexion gap for both the intact and the reconstructed knees closes with increasing flexion between 90° and 140° and opens up with further knee flexion. This in vitro study implies that the high flexion gap is an important concept that highly depends on the patient and on component design. Future investigations should include other designs as well as in vivo exploration to better understand the role of the high flexion gap in allowing patients to flex to deep angles.

7.5 References

1. Aglietti, P., et al., *An analysis of the survival rate of total-condylar total knee prostheses with posterior stability.* Ital J Orthop Traumatol, 1988. 14(4): p. 419-28.

2. Aglietti, P. and R. Buzzi, *Posteriorly stabilised total-condylar knee replacement. Three to eight years' follow-up of 85 knees.* J Bone Joint Surg Br, 1988. 70(2): p. 211-6.

3. Emmerson, K.P., C.G. Moran, and I.M. Pinder, *Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up.* J Bone Joint Surg Br, 1996. 78(3): p. 441-5.

4. Ranawat, C.S., et al., *Long-term results of the total condylar knee arthroplasty. A 15-year survivorship study.* Clin Orthop, 1993(286): p. 94-102.

5. Dennis, D., et al., *Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation.* Clin Orthop, 1992. 281: p. 168-176.

6. Insall, J.N., et al., *The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements.* J Bone Joint Surg Am, 1983. 65(5): p. 619-28.

7. Mangaleshkar, S.R., S.K. Bajaj, and A.P. Thomas, *Denham total knee arthroplasty: a 10-year follow-up study.* J Arthroplasty, 2002. 17(5): p. 550-5.

8. Khaw, F.M., L.M. Kirk, and P.J. Gregg, *Survival analysis of cemented Press-Fit Condylar total knee arthroplasty.* J Arthroplasty, 2001. 16(2): p. 161-7.

9. Ritter, M., et al., *Long-term survival analysis of a posterior cruciate-retaining total condylar total knee arthroplasty.* Clin Ortho, 1994. 309: p. 136-145.

10. Mulholland, S.J. and U.P. Wyss, *Activities of daily living in non-Western cultures: range of motion requirements for hip and knee joint implants.* Int J Rehabil Res, 2001. 24(3): p. 191-8.

11. Nagura, T., et al., *Mechanical loads at the knee joint during deep flexion.* J Orthop Res, 2002. 20(4): p. 881-6.

12. Laubenthal, K.N., G.L. Smidt, and D.B. Kettelkamp, *A quantitative analysis of knee motion during activities of daily living.* Phys Ther, 1972. 52(1): p. 34-43.

13. Walker, P.S. and A. Garg, *Range of motion in total knee arthroplasty. A computer analysis.* Clin Orthop, 1991(262): p. 227-35.

14. Rowe, P.J., et al., *Knee joint kinematics in gait and other functional activities measured using flexible electrogoniometry: how much knee motion is sufficient for normal daily life?* Gait Posture, 2000. 12(2): p. 143-55.

15. Li, G., et al., *Improving flexion in total knee arthroplasty*, in *The Adult Knee*, J. Callaghan, et al., Editors. 2002, Lippincott Williams & Wilkins: Philadelphia. p. 1233-1244.

16. Hefzy, M.S., B.P. Kelly, and T.D. Cooke, *Kinematics of the knee joint in deep flexion: a radiographic assessment.* Med Eng Phys, 1998. 20(4): p. 302-7.

17. Freeman, M.A. and G.T. Railton, *Should the posterior cruciate ligament be retained or resected in condylar nonmeniscal knee arthroplasty? The case for resection.* J Arthroplasty, 1988. 3 Suppl: p. S3-12.

18. Griffin, F.M., J.N. Insall, and G.R. Scuderi, *Accuracy of soft tissue balancing in total knee arthroplasty.* J Arthroplasty, 2000. 15(8): p. 970-3.

19. Insall, J.N., *Presidential address to The Knee Society. Choices and compromises in total knee arthroplasty.* Clin Orthop, 1988(226): p. 43-8.

20. Insall, J.N., *Technique of total knee replacement.* Insrt Course Lect, 1981. 30: p. 324.

21. Most, E., *Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement*, in *Department of Mechanical Engineering*. 2000, June, MIT: Cambridge.

22. Most, E., et al., *Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty.* Clin Orthop, 2003(410): p. 101-13.

23. Li, G., et al., *Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study.* Clin Orthop, 2002(404): p. 214-25.

24. Li, G., et al., *Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads.* J Arthroplasty, 2001. 16(8 Suppl 1): p. 150-6.

25. Most, E., et al., *The Kinematics of Fixed and Mobile Bearing Posterior Stabilized Total Knee Arthroplasty.* Clin Ortho, 2003. 416: p. 1-11.

26. Weiss, J.M., et al., *What functional activities are important to patients with knee replacements?* Clin Orthop, 2002(404): p. 172-88.

27. Ritter, M. and E. Stringer, *Predictive range of motion after total knee replacement.* Clin Orthop, 1979. 143: p. 115-119.

28. Parlsley, B., G. Engh, and K. Dwyer, *Preoperative flexion. Does it influence postoperative flexion after Posterior-Cruciate-Retaining total knee arthroplasty?* Clin Orthop, 1992. 275: p. 204-210.

29. Kawamura, H. and R. Bourne, *Factors affecting range of flexion after total knee arthroplasty.* J Orthop Sci, 2001. 6: p. 248-52.

30. Harvey, I., et al., *Factors affecting the range of movement of total knee arthroplasty.* J Bone Joint Surg, 1993. 75-B: p. 950-955.

31. Lizaur, A., L. Marco, and R. Cebrian, *Preoperative factors influencing the range of movement after Total Knee Arthroplasty for severe osteoarthritis.* J Bone Joint Surg Br, 1997. 79: p. 626-629.

32. Anouchi, Y., et al., *Range of motion in total knee replacement.* Clin Orthop, 1996. 331: p. 87-92.

33. Sultan, P., et al., *Optimizing Flexion After Total Knee Arthroplasty: Advances in Prosthetic Design.* Clin Orthop, 2003.

34. Andriacchi, T. and J. Galante, *Retention of the posterior cruciate in total knee arhtroplasty.* J Arthroplasty, 1988. Supplement: p. S13-S19.

35. Maloney, W.J. and D.J. Schurman, *The effects of implant design on range of motion after total knee arthroplasty. Total condylar versus posterior stabilized total condylar designs.* Clin Orthop, 1992(278): p. 147-52.

36. Chen, B., et al., *Continuous passive motion after total knee arthroplasty: a prospective study.* Am J Phys Med Rehabil, 2000. 79(5): p. 421-6.

37. Spanu, C. and M. Hefzy, *Biomechanics of the knee joint un deep felxion: A prelude to a total knee replacement that allow for maximum flexion.* Technology and Health Care, 2003. 11: p. 161-181.

38. Bellemans, J., et al., *Fluoroscopic analysis of the kinematics of deep flexion in total knee arthroplasty. Influence of posterior condylar offset.* J Bone Joint Surg Br, 2002. 84(1): p. 50-3.

39. Nakagawa, S., et al., *Tibiofemoral movement 3: full flexion in the living knee studied by MRI.* J Bone Joint Surg Br, 2000. 82(8): p. 1199-200.

40. Kadoya, Y., et al., *Effects of posterior cruciate ligament resection on the tibiofemoral joint gap.* Clin Orthop, 2001(391): p. 210-7.

Chapter 8

Conclusions

8.1 Accomplishments Reviewed

This work has examined the behavior of the intact and reconstructed (TKA) knees from 0° to 150° of flexion with the hope of obtaining a great understanding of the factors that limit knee motion at high flexion angles. Knee flexion is integral to function in many situations of every day life and the amount of knee flexion has been linked to functional outcome and activities of daily living [1]. In many situations, patients require flexion beyond 90° such as squatting and kneeling [2], using the bathtub [3, 4], or participating in variety of low impact sport activities [5]. The intact knee, in many individuals, is able of flexing up to 160 [6, 7]. Unfortunately, the flexion achieved after contemporary TKA rarely exceeds 120° [8-16]. Current prosthetic designs and surgical techniques do not meet the needs of all patients who require deep knee flexion for their daily activities. As part of this dissertation, two commonly used TKA designs were used to investigate the performance of these TKA on knee function: (1) posterior cruciate retaining TKA, whereby the posterior cruciate ligament is retained, and (2) posterior cruciate stabilizing TKA, whereby the posterior cruciate ligament is resected and substituted by a cam-spine mechanism.

The research performed as part of this dissertation explores the behavior of intact knees to better understand the promoting and limiting factors that guide the knee at high flexion angles. It proceeds by evaluating the performance of various TKA designs and comparing them to the behavior of the intact knee. The experiments carried out in this work utilize an in vitro robotic experimental system. Aside from the accuracy and the repeatability advantages, this robotic testing system allows for sequential tests (such as reconstructions) on the same specimen thereby limiting inter-specimen variations. Furthermore, in all experiments, the intact knee was tested first and served as a reference for all other tests.

The work in this thesis encompasses various facets that are thought to play an important role in limiting the range of flexion following a TKA. The six major findings of this work are reviewed below.

<u>Passive path kinematics:</u> In all experiments, the femoral intact knee center gradually translated posteriorly with increasing flexion, reaching a maximum value at 150°. Following any TKA, at full knee flexion, the femoral component was positioned slightly more posteriorly as compared to the intact knee. With increasing flexion to 30°, the femur center translated anteriorly. With further increased in knee flexion, the femur gradually translated posteriorly reaching a peak value at 150°. All TKA designs partially restored intact knee kinematics with some achieving up to 90% at 150° of knee flexion. In all knees, the lateral femoral condyle translation exceeded the translation of the medial

femoral condyle signifying the presence of internal tibial rotation. Internal tibial rotation increased with increasing flexion angles for all knees.

Parametric analysis of muscle loads: Throughout the various experiments the effect of different muscle loads on knee kinematics, for both the intact and the reconstructed knees, was investigated. The effect of muscle co-contraction on knee kinematics was explored. The limitation of the robotic testing system had to be taken into account when applying external loads, as the maximum payload of the robotic testing system is 150kg. We applied three different muscle loads: isolated 400N quadriceps, isolated 200N hamstrings, and combined 400N quadriceps / 200N hamstrings loads. The two isolated muscle loads represent the local extremes of this study while the combined load simulated a 2:1 co-contraction ratio as published in the literature [17, 18]. The isolated quadriceps load had a significant effect on knee kinematics at low flexion angles, causing the femur to translate more posteriorly and the tibia to rotate more internally as compared to the unloaded knee state. Maximum effect of the muscle load on knee kinematics was observed at 30° of knee flexion. With further increasing flexion angles beyond 30°, the effect of isolated quadriceps on knee kinematics diminished. Negligible effect was noted at deep knee flexion. The application of isolated hamstrings load at low flexion angles caused the femur to translate anteriorly and the tibia to rotate externally when compared to the unloaded knee. This effect was most noticeable at 30° of knee flexion. As with the isolated quadriceps load, isolated hamstrings load had minimal effect on knee kinematics for flexion angles greater than 30°. Little effect of isolated muscle load application on knee kinematics was noted at deep knee flexion. As would be expected, the combined muscle load results were situated between the two local extremes. It is concluded that muscle loads play a minimal role in affecting knee kinematics at maximal knee flexion. It is suggested that the knee is highly constrained in high flexion and that there is a strong resistance in high knee flexion. We attribute this phenomenon to the compression of the posterior soft tissue at high knee flexion.

The role of the anterior and posterior cruciate ligaments: This thesis investigated the role of the anterior (ACL) and posterior (PCL) cruciate ligaments on intact knee kinematics. Moreover, this work is the first to measure the forces in these ligaments from 0° to 150° of flexion. The results from this study show that both cruciate ligaments played a minimal role in restraining knee motion at high flexion angles. The ACL carried peak load around 30° of flexion, which is consistence with other published data [17]. The PCL carried highest load in the mid flexion range (90°), which is also consistent with other published data [19, 20].

Currently, surgeons approach the knee anteriorly in order to perform a TKA. In this surgical technique, the ACL is resected is all cases. Retaining or substituting the PCL remains the surgeon's choice. The effect of the PCL on knee kinematics in a cruciate retaining TKA have been quantified for flexion angles less than 120° [21, 22]. Up to this date, no information regarding the PCL forces in a PCL-retaining TKA has been reported. Furthermore, no information is known about the role of the PCL after PCL-retaining TKA in high knee flexion. The work done in this dissertation is the first to quantify the role of the PCL in PCL-retaining TKAs.

The in situ forces of the PCL in the PCL-retaining TKA closely followed the behavior of the intact PCL. The force increased with increasing flexion up to 90° but decreased beyond 90° of knee flexion. However, at 90° of flexion, the magnitude of the

PCL forces in intact knees was greater than that measured in the PCL-retaining TKA (45% lower for a conventional TKA 30% lower for a high flexion design). With further increase in knee flexion, the force in the PCL decreased reaching a minimal level at 150°. It is important to remember that the PCL is the primary constraint in posterior knee stability. Even though the data demonstrated that the PCL in a PCL-retaining TKA might not be loaded to the level of the intact knee, it did contribute to posterior femoral translation. It was noted by the fact that posterior femoral translation was further reduced in the TKA when the PCL was transected. It is concluded that PCL is an important structure in improving posterior femoral translation after PCL-retaining TKA.

The role of the cam-spine mechanism in posterior-stabilized TKA: In general, surgeons face the challenge of "perfectly" restoring the PCL tightness after TKA. Consequently, many surgeons feel that it is simply better to resect the PCL and substitute its function by a cam-spine mechanism. The data reported in this work is the first to experimentally quantify the role and forces of the cam-spine mechanism in a posterior-stabilized TKA. In this work, the femoral rollback in a PCL-stabilized TKA was correlated with the timing of the cam-spine engagement. Below 75°, little femoral translation was observed for the PCL-stabilized TKA as the femoral cam did not engage the tibial spine. The later was noted by low cam-spine contact forces. At approximately 75°, engagement of the cam-spine occurred at the same time when increase in posterior femoral translation was observed. The engagement continued until 135° of flexion. At that point, disengagement of the femoral cam from the tibial spine was noticed. Despite the separation of the cam-spine, posterior femoral translation continued to rise in the high flexion designs. We attribute this phenomenon to the compression of the posterior soft tissue at high knee flexion.

The contact mechanism in high knee flexion: In the intact knee, the medial meniscus remained attached to the tibial plateau throughout the entire range of knee motion. This provided an "extended" surface on the tibia that allows the femur to rollback over the tibia in a stable manner at high knee flexion. On the lateral side, at lower flexion angles (<120°), the lateral meniscus closely followed the tibial plateau. However, beyond 120°, the lateral meniscus elongates posteriorly on the tibial plateau, extending beyond the tibial surface to provide the stability of the knee at deep flexion. This unique function is, however, eliminated as part of the TKA procedure. Therefore, component design must compensate for the menisci deficiency.

Examining the kinematics behavior of several TKA designs revealed that, on average, most designs closely restored intact knee kinematics throughout the entire range of knee flexion. Examination strictly of kinematics did not disclose the differences among the various TKA designs. The reason that differences were not appreciated by kinematics is due to the fact that kinematics is but one aspect that contributes to the overall performance of a TKA. When the contact mechanics at high knee flexion was examined, the high flexion TKA designs differentiated themselves from the standard TKA designs. There was no significant difference in contact area between the conventional and the high flexion PCL-retaining TKA designs in the flexion range of 0° - 90°. The contact location for both TKA designs translated posteriorly with increasing flexion. At 120°, the conventional design reached the polyethylene edge while the high flexion design was positioned more posteriorly with a larger contact area. Point loading of the femoral condyles with the polyethylene edge for the conventional PCL-retaining TKA was observed at 120°, 135° and 150° of knee flexion. At 150°, the PCL-retaining TKA femoral condyles were no longer articulating with the polyethylene but rather articulating with the tibial bone. The high flexion PCL-retaining design did not reach the polyethylene edge until 150°. This increased knee stability by approximately 20° when compared to the conventional design. In evaluating the mechanisms that limit patients from achieving high knee flexion, it is necessary to examine the kinematics as well as the contact mechanism for each design. It is easy to be misled by observing good kinematics but inadequate contact, particularly at high knee flexion.

The role of the posterior soft tissue compression: Despite low PCL forces, disengagement of the cam-spine mechanism, or application of various muscle loads, the femur is all TKA designs continues to translated posteriorly at high knee flexion. This increase in rollback was accredited to the compression of the posterior soft tissue which includes the posterior capsule, menisci (only in the intact knee), hamstring muscles, skin, and fat. At high knee flexion, tissue gets compressed between the femoral shaft and the posterior edge of the tibia pushing the tibia anteriorly (or the femur posteriorly) promoting further femoral rollback.

8.2 Future Directions

Several other factors have been briefly investigated during the course of this work. The role of tibial tray placement on knee function was exmined. It appears that placing the tibial tray more posteriorly, when compared to the traditional anterior alignment, may provide a larger tibiofemoral contact area and perhaps a greater stability at high knee flexion. Implant sizing and placement are two factors that must be further investigated to better understand their effect on the stability of the knee at deep flexion.

The work in this thesis is the first to define and quantify the concept of high flexion gap. The initial work presented in Chapter 7 revealed that for the design and bony landmarks used in this study, the gap closed up with increasing flexion in the range of 90° to 140° in both the intact and the PCL-stabilized knees. Beyond 140°, the gap opened up again. This is likely due to soft tissue compression. The gap behavior is dependent on the patient and the component design. Therefore, future investigations should include other designs, particularly a PCL-retaining TKA, to evaluate the role of the partial and / or complete resection of the PCL on the gap dimensions. Moreover, in vivo analysis must be performed to better understand the role of the high flexion gap in allowing patients to flex to deep angles

The cam-spine mechanism plays a major role in prompting posterior femoral translation in the flexion range of 75° to 135°. This study did not investigate the effect of altering the location and shape of the tibial spine on the kinematics and contact mechanics. For example, it is important to investigate the effect of placing the tibial spine more anteriorly or more posteriorly on knee kinematics. Factors such as the engagement timing of the tibial spine with the femoral cam, contact location, and contact stress, should be further explored. It is suggested that a parametric analysis should be performed to explore the design criteria on knee function and the ability to allow patients to achieve higher knee flexion.

In recent years, patients express their desire to become more active and engage in activities that require high knee flexion. As a result, implant companies introduced a new type of component that potentially allows patients to safely flex to higher flexion angles. The high flexion femoral component introduces a new debate among orthopaedic surgeons. On one hand, the femoral condyles in the new design are elongated posteriorly by 2 mm to potentially provide higher stability at deep knee flexion. On the other hand, in order to fit the component on the femoral shaft, 2 mm of posterior condyle bone are removed. The decision between preserving more bone and elongating posterior femoral condyle remains in the surgeon's hand. As of this date, it is unknown which decision will benefit patients the most. It is necessary to objectively evaluate the two options.

A close examination of currently available total knee replacement components reveals that in general, all the implants look alike. All TKA are composed of three components: the femoral, tibial and polyethylene. The femoral and tibial components closely mimic the native knee geometry. On the other hand, the polyethylene liner does not closely mimic the tibial plateau. In the native knee, the lateral tibial plateau is convex and the medial tibial plateau is concave. In TKA, both the lateral and the medial side of the polyethylene are concave. This might explain the larger femoral "rollback" that is seen in the intact knees when compared to the reconstructed knees. It is my recommendation that the shape of the polyethylene liner should be investigated to determine its effect on knee kinematics and the ability of patient to flex beyond 120°.

Finally, it is apparent that the menisci are extremely valuable in guiding the knee and providing the necessary stability at high knee flexion. Until today, little was known about the performance of the menisci at high flexion angles. Unfortunately, the elimination of the menisci from the intact knee as part of the TKA procedure, presents a real challenge. None of the currently available TKA fully restores the behavior and the function of the intact knee at high knee flexion. It is my opinion that the concept of the menisci should be incorporated into the TKA design either via a material or design modifications. For example, altering the material or the composition of the polyethylene liner to enhance its properties (flexibility) while maintaining minimal wear behavior could be one such solution. A gel-like material might be one solution that would provide the necessary load distribution throughout the entire range of flexion while allowing the material to elongate over the tibial plateau similarly to the intact knee menisci. This is particularly important on the lateral side as it is shown that beyond 120°, the lateral meniscus elongates posteriorly on the tibial plateau, extending beyond the tibial surface to provide stability for the knee. Moreover, any other combination of gel-like material and ultra high polymethyl methacrylate may be used to balance between minimal wear and maximum flexion. Further investigation regarding the effect of the material on knee kinematics is necessary.

This thesis provides insightful information that will aid researchers in designing the next era in TKA.

8.3 References

1. Laubenthal, K.N., G.L. Smidt, and D.B. Kettelkamp, *A quantitative analysis of knee motion during activities of daily living.* Phys Ther, 1972. 52(1): p. 34-43.

2. Yamazaki, J., et al., *Hy-Flex II total knee system and range of motion.* Arch Orthop Trauma Surg, 2002. 122(3): p. 156-60.

3. Myles, C.M., et al., *Knee joint functional range of movement prior to and following total knee arthroplasty measured using flexible electrogoniometry.* Gait Posture, 2002. 16(1): p. 46-54.

4. Rowe, P.J., et al., *Knee joint kinematics in gait and other functional activities measured using flexible electrogoniometry: how much knee motion is sufficient for normal daily life?* Gait Posture, 2000. 12(2): p. 143-55.

5. Sultan, P., et al., *Optimizing Flexion After Total Knee Arthroplasty: Advances in Prosthetic Design.* Clin Orthop, 2003.

6. Mulholland, S.J. and U.P. Wyss, *Activities of daily living in non-Western cultures: range of motion requirements for hip and knee joint implants.* Int J Rehabil Res, 2001. 24(3): p. 191-8.

7. Hefzy, M.S., et al., *Knee kinematics in-vivo of kneeling in deep flexion examined by bi-planar radiographs.* Biomed Sci Instrum, 1997. 33: p. 453-8.

8. Aglietti, P., R. Buzzi, and A. Gaudenzi, *Patellofemoral functional results and complications with the posterior stabilized total condylar knee prosthesis.* J Arthroplasty, 1988. 3(1): p. 17-25.

9. Dennis, D., et al., *Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation.* Clin Orthop, 1992. 281: p. 168-176.

10. Emmerson, K.P., C.G. Moran, and I.M. Pinder, *Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up.* J Bone Joint Surg Br, 1996. 78(3): p. 441-5.

11. Goldberg, V.M., et al., *Use of a total condylar knee prosthesis for treatment of osteoarthritis and rheumatoid arthritis. Long-term results.* J Bone Joint Surg Am, 1988. 70(6): p. 802-11.

12. Insall, J.N., et al., *The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements.* J Bone Joint Surg Am, 1983. 65(5): p. 619-28.

13. Ranawat, C.S., C.P. Luessenhop, and J.A. Rodriguez, *The press-fit condylar modular total knee system. Four-to-six-year results with a posterior-cruciate-substituting design.* J Bone Joint Surg Am, 1997. 79(3): p. 342-8.

14. Ranawat, C.S., et al., *Long-term results of the total condylar knee arthroplasty. A 15-year survivorship study.* Clin Orthop, 1993(286): p. 94-102.

15. Rand, J.A., *Comparison of metal-backed and all-polyethylene tibial components in cruciate condylar total knee arthroplasty.* J Arthroplasty, 1993. 8(3): p. 307-13.

16. Rosenberg, A.G., R.M. Barden, and J.O. Galante, *Cemented and ingrowth fixation of the Miller-Galante prosthesis. Clinical and roentgenographic comparison after three- to six-year follow-up studies.* Clin Orthop, 1990(260): p. 71-9.

17. Li, G., et al., *The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL.* J Biomech, 1999. 32(4): p. 395-400.

18. Li, G., et al., *Prediction of antagonistic muscle forces using inverse dynamic optimization during flexion/extension of the knee.* J Biomech Eng, 1999. 121(3): p. 316-22.

19. Hoher, J., et al., *In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study.* J Orthop Res, 1999. 17(5): p. 763-8.

20. Zayontz, S., et al. *ACL and PCL do not carry significant loads in high flexion of the knee under muscle loads.* in *Trans Orthop Res So*. 2002. Dallas.

21. Stiehl, J. B., R. D. Komistek, et al. Clin Orthop 1999. "*Detrimental kinematics of a flat on flat total condylar knee arthroplasty.*" (365): p. 139-48.

22. Stiehl, J. B., R. D. Komistek, et al. J Bone Joint Surg Br 1995. "*Fluoroscopic analysis of kinematics after posterior-cruciate-retaining knee arthroplasty.*" 77(6): p. 884-9.

BIBLIOGRAPHY

Aglietti, P. and R. Buzzi (1988). "Posteriorly stabilised total-condylar knee replacement. Three to eight years' follow-up of 85 knees." J Bone Joint Surg Br 70(2): 211-6.

Aglietti, P., R. Buzzi, et al. (1988). "Patellofemoral functional results and complications with the posterior stabilized total condylar knee prosthesis." J Arthroplasty 3(1): 17-25.

Aglietti, P., F. Scrobe, et al. (1988). "An analysis of the survival rate of total-condylar total knee prostheses with posterior stability." Ital J Orthop Traumatol 14(4): 419-28.

Akagi, M., T. Nakamura, et al. (2000). "The Bisurface total knee replacement: a unique design for flexion. Four- to-nine-year follow-up study." J Bone Joint Surg Am 82-A(11): 1626-33.

Alexander, E. J. and T. P. Andriacchi (2001). "Correcting for deformation in skin-based marker systems." J Biomech 34(3): 355-61.

Andriacchi, T. and J. Galante (1988). "Retention of the posterior cruciate in total knee arhtroplasty." J Arthroplasty Supplement: S13-S19.

Andriacchi, T. P. (1993). "Functional analysis of pre and post-knee surgery: total knee arthroplasty and ACL reconstruction." J Biomech Eng 115(4B): 575-81.

Andriacchi, T. P., E. J. Alexander, et al. (1998). "A point cluster method for in vivo motion analysis: applied to a study of knee kinematics." J Biomech Eng 120(6): 743-9.

Andriacchi, T. P., J. O. Galante, et al. (1982). "The influence of total knee-replacement design on walking and stair-climbing." J Bone Joint Surg Am 64(9): 1328-35.

Anouchi, Y., M. McShane, et al. (1996). "Range of motion in total knee replacement." Clin Orthop 331: 87-92.

Arima, J., L. A. Whiteside, et al. (1998). "Effect of partial release of the posterior cruciate ligament in total knee arthroplasty." Clin Orthop(353): 194-202.

Arms, S. W., M. H. Pope, et al. (1984). "The biomechanics of anterior cruciate ligament rehabilitation and reconstruction." Am J Sports Med 12(1): 8-18.

Asano, T., M. Akagi, et al. (2001). "In vivo three-dimensional knee kinematics using a biplanar image- matching technique." Clin Orthop(388): 157-66.

Banks, S. A., G. D. Markovich, et al. (1997). "In vivo kinematics of cruciate-retaining and -substituting knee arthroplasties." J Arthroplasty 12(3): 297-304.

Bartel, D. L., A. H. Burstein, et al. (1985). "The effect of conformity and plastic thickness on contact stresses in metal-backed plastic implants." J. Biomech. Eng. 107: 193-199.

Bayley, J. C., R. D. Scott, et al. (1988). "Failure of the metal-backed patellar component after total knee replacement." J Bone Joint Surg Am 70(5): 668-74.

Bellemans, J., S. Banks, et al. (2002). "Fluoroscopic analysis of the kinematics of deep flexion in total knee arthroplasty. Influence of posterior condylar offset." J Bone Joint Surg Br 84(1): 50-3.

Berger, R. A., L. S. Crossett, et al. (1998). "Malrotation causing patellofemoral complications after total knee arthroplasty." Clin Orthop(356): 144-53.

Berger, R. A., H. E. Rubash, et al. (1993). "Determining the rotational alignment of the femoral component in total knee arthroplasty using the epicondylar axis." Clin Orthop(286): 40-7.

Beynnon, B. D., B. C. Fleming, et al. (1995). "Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo." Am J Sports Med 23(1): 24-34.

Blankevoort, L., R. Huiskes, et al. (1988). "The envelope of passive knee joint motion." J Biomech 21(9): 705-20.

Blankevoort, L., R. Huiskes, et al. (1990). "Helical axes of passive knee joint motions." J Biomech 23(12): 1219-29.

Bryan, R. S. and J. A. Rand (1982). "Revision total knee arthroplasty." Clin Orthop(170): 116-22.

Buechel, F. F., Sr., F. F. Buechel, Jr., et al. (2001). "Twenty-year evaluation of meniscal bearing and rotating platform knee replacements." Clin Orthop 388: 41-50.

Butler, D. L., F. R. Noyes, et al. (1980). "Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study." J Bone Joint Surg Am 62(2): 259-70.

Callaghan, J., J. Insall, et al. (2000). "Mobile-Bearing knee replacement: concepts and results." J Bone Joint Surg Am 82(7): 1020.

Callaghan, J. J., M. W. Squire, et al. (2000). "Cemented rotating-platform total knee replacement. A nine to twelve- year follow-up study." J Bone Joint Surg Am 82(5): 705-11.

Chen, B., J. R. Zimmerman, et al. (2000). "Continuous passive motion after total knee arthroplasty: a prospective study." Am J Phys Med Rehabil 79(5): 421-6.

Churchill, D. L., S. J. Incavo, et al. (1998). "The transepicondylar axis approximates the optimal flexion axis of the knee." Clin Orthop(356): 111-8.

Colizza, W. A., J. N. Insall, et al. (1995). "The posterior stabilized total knee prosthesis. Assessment of polyethylene damage and osteolysis after a ten-year-minimum follow-up." J Bone Joint Surg Am 77(11): 1713-20.

Collier, J. P., J. L. McNamara, et al. (1991). "All-polyethylene patellar components are not the answer." Clin Orthop(273): 198-203.

de Lange, A., R. Huiskes, et al. (1990). "Measurement errors in roentgen-stereophotogrammetric joint-motion analysis." J Biomech 23(3): 259-69.

Delp, S. L., J. H. Kocmond, et al. (1995). "Tradeoffs between motion and stability in posterior substituting knee arthroplasty design." J Biomech 28(10): 1155-66.

Dennis, D., M. Clayton, et al. (1992). "Posterior cruciate condylar total knee arthroplasty: average 11-year followup evaluation." Clin Orthop 281: 168-176.

Dennis, D., R. Komistek, et al. (1997). "In Vivo femoral condylar lift-off in total knee arthroplasty." Orthop Trans 21: 1112.

Dennis, D. A., R. D. Komistek, et al. (1998). "In vivo anteroposterior femorotibial translation of total knee arthroplasty: a multicenter analysis." Clin Orthop(356): 47-57.

Dennis, D. A., R. D. Komistek, et al. (1996). "In vivo knee kinematics derived using an inverse perspective technique." Clin Orthop(331): 107-17.

Dennis, D. A., R. D. Komistek, et al. (2003). "In vivo fluoroscopic analysis of fixed-bearing total knee replacements." Clin Orthop(410): 114-30.

Dennis, D. A., R. D. Komistek, et al. (1998). "Range of motion after total knee arthroplasty: the effect of implant design and weight-bearing conditions." J Arthroplasty 13(7): 748-52.

DesJardins, J. D., P. S. Walker, et al. (2000). "The use of a force-controlled dynamic knee simulator to quantify the mechanical performance of total knee replacement designs during functional activity." J Biomech 33(10): 1231-42.

D'Lima, D. D., M. Trice, et al. (2000). "Comparison between the kinematics of fixed and rotating bearing knee prostheses." Clin Orthop(380): 151-7.

Draganich, L. F., T. P. Andriacchi, et al. (1987). "Interaction between intrinsic knee mechanics and the knee extensor mechanism." J Orthop Res 5(4): 539-47.

Duffy, G. P., R. T. Trousdale, et al. (1998). "Total knee arthroplasty in patients 55 years old or younger. 10- to 17-year results." Clin Orthop(356): 22-7.

Durselen, L., L. Claes, et al. (1995). "The influence of muscle forces and external loads on cruciate ligament strain." Am J Sports Med 23(1): 129-36.

Eckhoff, D. G., T. F. Dwyer, et al. (2001). "Three-dimensional morphology of the distal part of the femur viewed in virtual reality." J Bone Joint Surg Am 83-A(Suppl 2(Pt 1)): 43-50.

Emmerson, K. P., C. G. Moran, et al. (1996). "Survivorship analysis of the Kinematic Stabilizer total knee replacement: a 10- to 14-year follow-up." J Bone Joint Surg Br 78(3): 441-5.

Essinger, J. R., P. F. Leyvraz, et al. (1989). "A mathematical model for the evaluation of the behaviour during flexion of condylar-type knee prostheses." J Biomech 22(11-12): 1229-41.

Fanelli, G. C., B. F. Giannotti, et al. (1994). "The posterior cruciate ligament arthroscopic evaluation and treatment." Arthroscopy 10(6): 673-88.

Feikes, J., D. Wilson, et al. (1998). The Unique Track of Intact Passive Knee Motion as a Kinematic Baseline. Orthopaedic Research Society, New Orleans, Louisiana.

Fox, R. J., C. D. Harner, et al. (1998). "Determination of the in situ forces in the human posterior cruciate ligament using robotic technology. A cadaveric study." Am J Sports Med 26(3): 395-401.

Freeman, M. A. and V. Pinskerova (2003). "The movement of the knee studied by magnetic resonance imaging." Clin Orthop(410): 35-43.

Freeman, M. A. and G. T. Railton (1988). "Should the posterior cruciate ligament be retained or resected in condylar nonmeniscal knee arthroplasty? The case for resection." J Arthroplasty 3 Suppl: S3-12.

Fujie, H., G. A. Livesay, et al. (1996). "Forces and moments in six-DOF at the human knee joint: mathematical description for control." J Biomech 29(12): 1577-85.

Fujie, H., G. A. Livesay, et al. (1995). "The use of a universal force-moment sensor to determine in-situ forces in ligaments: a new methodology." J Biomech Eng 117(1): 1-7.

Fujie, H., K. Mabuchi, et al. (1993). "The use of robotics technology to study human joint kinematics: a new methodology." J Biomech Eng 115(3): 211-7.

Gill, G. S. and A. B. Joshi (2001). "Long-term results of retention of the posterior cruciate ligament in total knee replacement in rheumatoid arthritis." J Bone Joint Surg Br 83(4): 510-2.

Gill, T. J., L. E. DeFrate, et al. (2003). "The biomechanical effect of posterior cruciate ligament reconstruction on knee joint function. Kinematic response to simulated muscle loads." Am J Sports Med 31(4): 530-6.

Godest, A. C., C. S. de Cloke, et al. (2000). "A computational model for the prediction of total knee replacement kinematics in the sagittal plane." J Biomech 33(4): 435-42.

Goldberg, V. M., M. P. Figgie, et al. (1988). "Use of a total condylar knee prosthesis for treatment of osteoarthritis and rheumatoid arthritis. Long-term results." J Bone Joint Surg Am 70(6): 802-11.

Gollehon, D. L., P. A. Torzilli, et al. (1987). "The role of the posterolateral and cruciate ligaments in the stability of the human knee. A biomechanical study." J Bone Joint Surg Am 69(2): 233-42.

Goodfellow, J. and J. O'Connor (1978). "The mechanics of the knee and prosthesis design." J Bone Joint Surg Br 60-B(3): 358-69.

Griffin, F. M., J. N. Insall, et al. (2000). "Accuracy of soft tissue balancing in total knee arthroplasty." J Arthroplasty 15(8): 970-3.

Grood, E. S. and W. J. Suntay (1983). "A joint coordinate system for the clinical description of three-dimensional motions: application to the knee." J Biomech Eng 105(2): 136-44.

Harner, C. D., T. M. Vogrin, et al. (2000). "Biomechanical analysis of a posterior cruciate ligament reconstruction. Deficiency of the posterolateral structures as a cause of graft failure." Am J Sports Med 28(1): 32-9.

Harvey, I., K. Barry, et al. (1993). "Factors affecting the range of movement of total knee arthroplasty." J Bone Joint Surg 75-B: 950-955.

Hefzy, M. S., B. P. Kelly, et al. (1998). "Kinematics of the knee joint in deep flexion: a radiographic assessment." Med Eng Phys 20(4): 302-7.

Hefzy, M. S., B. P. Kelly, et al. (1997). "Knee kinematics in-vivo of kneeling in deep flexion examined by bi-planar radiographs." Biomed Sci Instrum 33: 453-8.

Hill, P. F., V. Vedi, et al. (2000). "Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI." J Bone Joint Surg Br 82(8): 1196-8.

Hirokawa, S., M. Solomonow, et al. (1992). "Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction." Am J Sports Med 20(3): 299-306.

Hoher, J., T. M. Vogrin, et al. (1999). "In situ forces in the human posterior cruciate ligament in response to muscle loads: a cadaveric study." J Orthop Res 17(5): 763-8.

Hollister, A. M., S. Jatana, et al. (1993). "The axes of rotation of the knee." Clin Orthop(290): 259-68.

Insall, J. N. (1981). "Technique of total knee replacement." Insrt Course Lect 30: 324.

Insall, J. N. (1988). "Presidential address to The Knee Society. Choices and compromises in total knee arthroplasty." Clin Orthop(226): 43-8.

Insall, J. N., R. Binazzi, et al. (1985). "Total knee arthroplasty." Clin Orthop(192): 13-22.

Insall, J. N., R. W. Hood, et al. (1983). "The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements." J Bone Joint Surg Am 65(5): 619-28.

Insall, J. N., P. F. Lachiewicz, et al. (1982). "The posterior stabilized condylar prosthesis: a modification of the total condylar design. Two to four-year clinical experience." J Bone Joint Surg Am 64(9): 1317-23.

Iwaki, H., V. Pinskerova, et al. (2000). "Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee." J Bone Joint Surg Br 82(8): 1189-95.

Kadoya, Y., A. Kobayashi, et al. (2001). "Effects of posterior cruciate ligament resection on the tibiofemoral joint gap." Clin Orthop(391): 210-7.

Kanamori, A., S. L. Woo, et al. (2000). "The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology." Arthroscopy 16(6): 633-9.

Kaper, B. P., P. N. Smith, et al. (1999). "Medium-term results of a mobile bearing total knee replacement." Clin Orthop(367): 201-9.

Kawamura, H. and R. Bourne (2001). "Factors affecting range of flexion after total knee arthroplasty." J Orthop Sci 6: 248-52.

Khaw, F. M., L. M. Kirk, et al. (2001). "Survival analysis of cemented Press-Fit Condylar total knee arthroplasty." J Arthroplasty 16(2): 161-7.

Kim, H., R. R. Pelker, et al. (1997). "Rollback in posterior cruciate ligament-retaining total knee arthroplasty. A radiographic analysis." J Arthroplasty 12(5): 553-61.

Kinzel, G. L., A. S. Hall, Jr., et al. (1972). "Measurement of the total motion between two body segments. I. Analytical development." J Biomech 5(1): 93-105.

Kinzel, G. L., B. M. Hillberry, et al. (1972). "Measurement of the total motion between two body segments. II. Description of application." J Biomech 5(3): 283-93.

Kirstukas, S. J., J. L. Lewis, et al. (1992). "6R instrumented spatial linkages for anatomical joint motion measurement--Part 1: Design." J Biomech Eng 114(1): 92-100.

Kirstukas, S. J., J. L. Lewis, et al. (1992). "6R instrumented spatial linkages for anatomical joint motion measurement--Part 2: Calibration." J Biomech Eng 114(1): 101-10.

Komistek, R. D., D. A. Dennis, et al. (2003). "In vivo fluoroscopic analysis of the normal human knee." Clin Orthop(410): 69-81.

Komistek, R. D., R. D. Scott, et al. (2002). "In vivo comparison of femorotibial contact positions for press-fit posterior stabilized and posterior cruciate-retaining total knee arthroplasties." J Arthroplasty 17(2): 209-16.

Koshino, T., T. Saito, et al. (2002). "Increase in range of knee motion to obtain floor sitting after high tibial osteotomy for osteoarthritis." Knee 9(3): 189-96.

Lafortune, M. A., P. R. Cavanagh, et al. (1992). "Three-dimensional kinematics of the human knee during walking." J Biomech 25(4): 347-57.

Laskin, R. S. and H. M. O'Flynn (1997). "The Insall Award. Total knee replacement with posterior cruciate ligament retention in rheumatoid arthritis. Problems and complications." Clin Orthop(345): 24-8.

Laubenthal, K. N., G. L. Smidt, et al. (1972). "A quantitative analysis of knee motion during activities of daily living." Phys Ther 52(1): 34-43.

Lee, J. G., E. M. Keating, et al. (1990). "Review of the all-polyethylene tibial component in total knee arthroplasty. A minimum seven-year follow-up period." Clin Orthop(260): 87-92.

Lewandowski, P. J., M. J. Askew, et al. (1997). "Kinematics of posterior cruciate ligament-retaining and -sacrificing mobile bearing total knee arthroplasties. An in vitro comparison of the New Jersey LCS meniscal bearing and rotating platform prostheses." J Arthroplasty 12(7): 777-84.

Lewis, J. L., W. D. Lew, et al. (1988). "Description and error evaluation of an in vitro knee joint testing system." J Biomech Eng 110(3): 238-48.

Li, G., L. DeFrate, et al. (2003). "Determination of optimal graft lengths for posterior cruciate ligament reconstruction--a theoretical analysis." J Biomech Eng 125(2): 295-9.

Li, G., T. J. Gill, et al. (2002). "Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads--an in vitro experimental study." J Orthop Res 20(4): 887-92.

Li, G., K. R. Kaufman, et al. (1999). "Prediction of antagonistic muscle forces using inverse dynamic optimization during flexion/extension of the knee." J Biomech Eng 121(3): 316-22.

Li, G., E. Most, et al. (2002). "Biomechanics of posterior-substituting total knee arthroplasty: an in vitro study." Clin Orthop(404): 214-25.

Li, G., T. W. Rudy, et al. (1998). "Effect of combined axial compressive and anterior tibial loads on in situ forces in the anterior cruciate ligament: a porcine study." J Orthop Res 16(1): 122-7.

Li, G., T. W. Rudy, et al. (1999). "The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL." J Biomech 32(4): 395-400.

Li, G., S. L. Schule, et al. (2002). Improving flexion in total knee arthroplasty. The Adult Knee. J. Callaghan, A. G. Rosenberg, H. E. Rubash, P. T. Simonian and T. A. Wickiewicz. Philadelphia, Lippincott Williams & Wilkins. II: 1233-1244.

Li, G., S. Zayontz, et al. (2001). "Cruciate-retaining and cruciate-substituting total knee arthroplasty: an in vitro comparison of the kinematics under muscle loads." J Arthroplasty 16(8 Suppl 1): 150-6.

Livesay, G. A., T. W. Rudy, et al. (1997). "Evaluation of the effect of joint constraints on the in situ force distribution in the anterior cruciate ligament." J Orthop Res 15(2): 278-84.

Lizaur, A., L. Marco, et al. (1997). "Preoperative factors influencing the range of movement after Total Knee Arthroplasty for severe osteoarthritis." J Bone Joint Surg Br 79: 626-629.

Lombardi, A. V., Jr., G. A. Engh, et al. (1988). "Fracture/dissociation of the polyethylene in metal-backed patellar components in total knee arthroplasty." J Bone Joint Surg Am 70(5): 675-9.

Lucchetti, L., A. Cappozzo, et al. (1998). "Skin movement artefact assessment and compensation in the estimation of knee-joint kinematics." J Biomech 31(11): 977-84.

Mahoney, O. M., P. C. Noble, et al. (1994). "Posterior cruciate function following total knee arthroplasty. A biomechanical study." J Arthroplasty 9(6): 569-78.

Malkani, A. L., J. A. Rand, et al. (1995). "Total knee arthroplasty with the kinematic condylar prosthesis. A ten- year follow-up study." J Bone Joint Surg Am 77(3): 423-31.

Mallon, W. J. and J. J. Callaghan (1993). "Total knee arthroplasty in active golfers." J Arthroplasty 8(3): 299-306.

Maloney, W. J. and D. J. Schurman (1992). "The effects of implant design on range of motion after total knee arthroplasty. Total condylar versus posterior stabilized total condylar designs." Clin Orthop(278): 147-52.

Mangaleshkar, S. R., S. K. Bajaj, et al. (2002). "Denham total knee arthroplasty: a 10-year follow-up study." J Arthroplasty 17(5): 550-5.

Maquet, P. (1976). "Advancement of the tibial tuberosity." Clin Orthop(115): 225-30.

Maquet, P. (1979). "Mechanics and osteoarthritis of the patellofemoral joint." Clin Orthop(144): 70-3.

Markolf, K. L., D. M. Burchfield, et al. (1996). "Biomechanical consequences of replacement of the anterior cruciate ligament with a patellar ligament allograft. Part II: forces in the graft compared with forces in the intact ligament." J Bone Joint Surg Am 78(11): 1728-34.

Markolf, K. L., D. M. Burchfield, et al. (1996). "Biomechanical consequences of replacement of the anterior cruciate ligament with a patellar ligament allograft. Part I: insertion of the graft and anterior-posterior testing." J Bone Joint Surg Am 78(11): 1720-7.

Markolf, K. L., J. R. Slauterbeck, et al. (1997). "A biomechanical study of replacement of the posterior cruciate ligament with a graft. Part 1: Isometry, pre-tension of the graft, and anterior-posterior laxity." J Bone Joint Surg Am 79(3): 375-80.

Markolf, K. L., J. R. Slauterbeck, et al. (1997). "A biomechanical study of replacement of the posterior cruciate ligament with a graft. Part II: Forces in the graft compared with forces in the intact ligament." J Bone Joint Surg Am 79(3): 381-6.

Matsuda, S., H. Miura, et al. (1999). "Knee stability in posterior cruciate ligament retaining total knee arthroplasty." Clin Orthop(366): 169-73.

Matsuda, S., S. E. White, et al. (1998). "Contact stress analysis in meniscal bearing total knee arthroplasty." J Arthroplasty 13(6): 699-706.

Meding, J. B., M. A. Ritter, et al. (2001). "Total knee arthroplasty with 4.4 mm of tibial polyethylene: 10-year followup." Clin Orthop(388): 112-7.

Miller, M. C., A. X. Zhang, et al. (2001). "The effect of component placement on knee kinetics after arthroplasty with an unconstrained prosthesis." J Orthop Res 19(4): 614-20.

Miller, R. K., J. W. Goodfellow, et al. (1998). "In vitro measurement of patellofemoral force after three types of knee replacement." J Bone Joint Surg Br 80(5): 900-6.

Miller, R. K., D. W. Murray, et al. (1997). "In vitro patellofemoral joint force determined by a non-invasive technique." Clin Biomech (Bristol, Avon) 12(1): 1-7.

Most, E. (2000, June). Development of a 6-DOF Robotic Test System for Studying the Biomechanics of Total Knee Replacement. Department of Mechanical Engineering. Cambridge, MIT.

Most, E., G. Li, et al. (2003). "The Kinematics of Fixed and Mobile Bearing Posterior Stabilized Total Knee Arthroplasty." Clin Ortho 416: 1-11.

Most, E., S. Zayontz, et al. (2003). "Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty." Clin Orthop(410): 101-13.

Mulholland, S. J. and U. P. Wyss (2001). "Activities of daily living in non-Western cultures: range of motion requirements for hip and knee joint implants." Int J Rehabil Res 24(3): 191-8.

Myles, C. M., P. J. Rowe, et al. (2002). "Knee joint functional range of movement prior to and following total knee arthroplasty measured using flexible electrogoniometry." Gait Posture 16(1): 46-54.

Nagura, T., C. O. Dyrby, et al. (2002). "Mechanical loads at the knee joint during deep flexion." J Orthop Res 20(4): 881-6.

Nakagawa, S., Y. Kadoya, et al. (2000). "Tibiofemoral movement 3: full flexion in the living knee studied by MRI." J Bone Joint Surg Br 82(8): 1199-200.

Olcott, C. W. and R. D. Scott (1999). "The Ranawat Award. Femoral component rotation during total knee arthroplasty." Clin Orthop(367): 39-42.

Parlsley, B., G. Engh, et al. (1992). "Preoperative flexion. Does it influence postoperative flexion after Posterior-Cruciate-Retaining total knee arthroplasty?" Clin Orthop 275: 204-210.

Pavone, V., F. Boettner, et al. (2001). "Total condylar knee arthroplasty: a long-term followup." Clin Orthop(388): 18-25.

Pereira, D. S., F. F. Jaffe, et al. (1998). "Posterior cruciate ligament-sparing versus posterior cruciate ligament-sacrificing arthroplasty. Functional results using the same prosthesis." J Arthroplasty 13(2): 138-44.

Piazza, S. J. and P. R. Cavanagh (2000). "Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment." J Biomech 33(8): 1029-34.

Piazza, S. J., S. L. Delp, et al. (1998). "Posterior tilting of the tibial component decreases femoral rollback in posterior-substituting knee replacement: a computer simulation study." J Orthop Res 16(2): 264-70.

Ranawat, C. S., W. F. Flynn, Jr., et al. (1993). "Long-term results of the total condylar knee arthroplasty. A 15-year survivorship study." Clin Orthop(286): 94-102.

Ranawat, C. S., C. P. Luessenhop, et al. (1997). "The press-fit condylar modular total knee system. Four-to-six-year results with a posterior-cruciate-substituting design." J Bone Joint Surg Am 79(3): 342-8.

Rand, J. A. (1993). "Comparison of metal-backed and all-polyethylene tibial components in cruciate condylar total knee arthroplasty." J Arthroplasty 8(3): 307-13.

Reinschmidt, C., A. J. van den Bogert, et al. (1997). "Effect of skin movement on the analysis of skeletal knee joint motion during running." J Biomech 30(7): 729-32.

Ritter, M., S. Herbst, et al. (1994). "Long-term survival analysis of a posterior cruciate-retaining total condylar total knee arthroplasty." Clin Ortho 309: 136-145.

Ritter, M. and E. Stringer (1979). "Predictive range of motion after total knee replacement." Clin Orthop 143: 115-119.

Ritter, M. A., P. M. Faris, et al. (1988). "Posterior cruciate ligament balancing during total knee arthroplasty." J Arthroplasty 3(4): 323-6.

Rosenberg, A. G., R. M. Barden, et al. (1990). "Cemented and ingrowth fixation of the Miller-Galante prosthesis. Clinical and roentgenographic comparison after three-to six-year follow-up studies." Clin Orthop(260): 71-9.

Rowe, P. J., C. M. Myles, et al. (2000). "Knee joint kinematics in gait and other functional activities measured using flexible electrogoniometry: how much knee motion is sufficient for normal daily life?" Gait Posture 12(2): 143-55.

Rudy, T. W., G. A. Livesay, et al. (1996). "A combined robotic/universal force sensor approach to determine in situ forces of knee ligaments." J Biomech 29(10): 1357-60.

Sakane, M., R. J. Fox, et al. (1997). "In situ forces in the anterior cruciate ligament and its bundles in response to anterior tibial loads." J Orthop Res 15(2): 285-93.

Sakane, M., G. A. Livesay, et al. (1999). "Relative contribution of the ACL, MCL, and bony contact to the anterior stability of the knee." Knee Surg Sports Traumatol Arthrosc 7(2): 93-7.

Schai, P. A., T. S. Thornhill, et al. (1998). "Total knee arthroplasty with the PFC system. Results at a minimum of ten years and survivorship analysis." J Bone Joint Surg Br 80(5): 850-8.

Silver, F., G. Bradica, et al. (2003). Structure and Biomechanics of Articular Cartilage. The Adult Knee. J. Callaghan, A. Rosenberg, H. Rubash, P. Simonian and T. Wickiewicz. Philadelphia, Lippincott Williams & Wilkins. 1: 105-122.

Singerman, R., J. C. Dean, et al. (1996). "Decreased posterior tibial slope increases strain in the posterior cruciate ligament following total knee arthroplasty." J Arthroplasty 11(1): 99-103.

Spanu, C. and M. Hefzy (2003). "Biomechanics of the knee joint un deep felxion: A prelude to a total knee replacement that allow for maximum flexion." Technology and Health Care 11: 161-181.

Stern, S. H. and J. N. Insall (1992). "Posterior stabilized prosthesis. Results after follow-up of nine to twelve years." J Bone Joint Surg Am 74(7): 980-6.

Stiehl, J. B. (2002). "World Experience with low Contact Stress mobile bearing Total Knee Arthroplasty: A literature Review." Orthopedics February, Supplement.

Stiehl, J. B., D. A. Dennis, et al. (1997). "In vivo kinematic analysis of a mobile bearing total knee prosthesis." Clin Orthop(345): 60-6.

Stiehl, J. B., D. A. Dennis, et al. (2000). "In vivo kinematic comparison of posterior cruciate ligament retention or sacrifice with a mobile bearing total knee arthroplasty." Am J Knee Surg 13(1): 13-8.

Stiehl, J. B., R. D. Komistek, et al. (1999). "Detrimental kinematics of a flat on flat total condylar knee arthroplasty." Clin Orthop(365): 139-48.

Stiehl, J. B., R. D. Komistek, et al. (1995). "Fluoroscopic analysis of kinematics after posterior-cruciate-retaining knee arthroplasty." J Bone Joint Surg Br 77(6): 884-9.

Sultan, P., E. Most, et al. (2003). "Optimizing Flexion After Total Knee Arthroplasty: Advances in Prosthetic Design." Clin Orthop.

Tarnowski, L. E., T. P. Andriacchi, et al. (1998). "Three-dimensional motion of cruciate retaining and posterior stabilized TKAs during walking." Trans ORS 44: 804.

Todo, S., Y. Kadoya, et al. (1999). "Anteroposterior and rotational movement of femur during knee flexion." Clin Orthop(362): 162-70.

Torzilli, P. A., X. Deng, et al. (1994). "The effect of joint-compressive load and quadriceps muscle force on knee motion in the intact and anterior cruciate ligament-sectioned knee." Am J Sports Med 22(1): 105-12.

Trousdale, R., T. Thornhill, et al. (2002). "Case Challenges in Hip & Knee Surgery: Knee Challenges: What Would You Do?" Orthopedics 25(9).

van Dijk, R., R. Huiskes, et al. (1979). "Roentgen stereophotogrammetric methods for the evaluation of the three dimensional kinematic behaviour and cruciate ligament length patterns of the human knee joint." J Biomech 12(9): 727-31.

Vertullo, C. J., M. E. Easly, et al. (2001). "Mobile Bearing In Primary Knee Arthroplasty." Journal of American Academy of orthopedic Surgeons 9(6): 355-64.

Vogrin, T. M., J. Hoher, et al. (2000). "Effects of sectioning the posterolateral structures on knee kinematics and in situ forces in the posterior cruciate ligament." Knee Surg Sports Traumatol Arthrosc 8(2): 93-8.

Walker, P. S. and A. Garg (1991). "Range of motion in total knee arthroplasty. A computer analysis." Clin Orthop(262): 227-35.

Wasielewski, R. C., N. Parks, et al. (1997). "Tibial insert undersurface as a contributing source of polyethylene wear debris." Clin Orthop(345): 53-9.

Weiss, J. M., P. C. Noble, et al. (2002). "What functional activities are important to patients with knee replacements?" Clin Orthop(404): 172-88.

Werner, F., D. Foster, et al. (1978). "The influence of design on the transmission of torque across knee prostheses." JBJS 60A: 342-348.

Wilson, D. R., J. D. Feikes, et al. (2000). "The components of passive knee movement are coupled to flexion angle." J Biomech 33(4): 465-73.

Wimmer, M. A. and T. P. Andriacchi (1997). "Tractive forces during rolling motion of the knee: implications for wear in total knee replacement." J Biomech 30(2): 131-7.

Worland, R. L., D. E. Jessup, et al. (1997). "Posterior cruciate recession in total knee arthroplasty." J Arthroplasty 12(1): 70-3.

Yamazaki, J., S. Ishigami, et al. (2002). "Hy-Flex II total knee system and range of motion." Arch Orthop Trauma Surg 122(3): 156-60.

Yoshino, N., S. Takai, et al. (2001). "Computed tomography measurement of the surgical and clinical transepicondylar axis of the distal femur in osteoarthritic knees." J Arthroplasty 16(4): 493-7.

Zavatsky, A. B. (1997). "A kinematic-freedom analysis of a flexed-knee-stance testing rig." J Biomech 30(3): 277-80.

Zayontz, S., E. Most, et al. (2002). ACL and PCL do not carry significant loads in high flexion of the knee under muscle loads. Trans Orthop Res So, Dallas.

Zimmer, I. NexGen LPS-Flex Design Rationale. Warsaw, IN.